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This research assessed changes in lower extremity biomechanics as a result of augmented local abdominal contractions during double leg landings. The study design followed a two-day (control and intervention days) within subject model in which two conditions on each day were compared, (control-control and control-experimental) with between sex comparisons. Fifty subjects (25 males and 25 females) were activity matched and represented a sample of healthy and recreationally active individuals.

A repeated measures ANOVA comparing control and experimental conditions on the intervention day revealed that all subjects significantly increased local abdominal activation during 150ms prior to landing. However, a 2 (sex) x 3 (muscle) x 2 (phase of landing) repeated measures ANOVA demonstrated that only males maintained this contraction during the 150ms time interval after landing. A repeated measures ANOVA evaluating changes in leg spring stiffness (LSS) as a result of the augmented local abdominal contraction showed a sex by condition interaction demonstrating that only males experienced significant increases in LSS from control to experimental condition while females demonstrated no significant changes. No significant differences by condition were noted when assessing changes in ankle, knee, and hip energetics. The evaluation of lower extremity total joint displacements across condition demonstrated that males increased LSS through decreases in hip range of motion while females showed no significant changes in hip range of motion but increased knee and ankle motion. From

these results we concluded that augmented local abdominal activation during a double leg landing task increased LSS by decreasing hip range of motion in males.

EFFECTS OF AUGMENTED LOCAL ABDOMINAL ACTIVATION PATTERNS ON LOWER EXTREMITY BIOMECHANICS DURING LANDING IN MALES AND FEMALES

by

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Approved by

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Committee Chair

To my wife Julie,

Your love and support made this accomplishment possible.

APPROVAL PAGE

This dissertation has been approved by the following committee of the Faculty of The Graduate School at the University of North Carolina at Greensboro.

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CHAPTER I

INTRODUCTION

In an effort to better understand postural faults and mechanisms that may predispose one to injury at the ankle, knee, or hip, the lower extremity is often assessed as a kinetic chain involving a series of interdependent joints. For example, based on clinically relevant postural assessments, Kendall et al (1993) considered that pelvic posture and/or motion can influence the rotations of the femur and thus affect the knee joint (Kendall, McCreary, & Provance, 1993). From the distal aspect of the kinetic chain, excessive pronation or supination at the foot can influence rotations of the tibia, and therefore influence knee mechanics (Kendall et al., 1993). Based on video and first-hand observations of knee injuries, Mary Lloyd Ireland termed the “Position of No Return” to describe a combination of postures within the kinetic chain that potentially contribute to knee injury, particularly the Anterior Cruciate Ligament (ACL) (Ireland, Gaudette, & Crook, 1997). These postures involve the foot fixed to the ground with excessive pronation, external rotation at the tibia, internal rotation of the femur, hip adduction, and the collective head, arms, and trunk (HAT) segment excessively forward flexed such that its mass is seemingly close to the outer limits of the base of support.

As rates of ACL injuries have been demonstrated to be higher in female athletes, laboratory research assessing the knee and its role within the kinetic chain has developed. Laboratory measures used to evaluate relationships between the lower extremity joints and the body as a whole include leg spring stiffness (LSS) and energy absorption (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Devita & Skelly, 1992; Farley, Houdijk, Strien, & Louie, 1998b; Granata, Padua, & Wilson, 2002). Landing models are commonly used to assess these measures as landing has been associated with lower extremity injuries (Boden, Dean, Feagin, & Garrett, 2000; Kirkendall & Garrett, 2000). While LSS reflects the ability of the lower extremity to decelerate the body's vertical momentum, energy absorption measures the individual joint contributions to attenuate the ground reaction forces at landing (Farley et al., 1998b; Schot & Dufek, 1993). Males have produced higher LSS than females (before using body mass as a covariate) during hopping tasks, while females have demonstrated higher knee and ankle energy absorption characteristics in landing (Decker et al., 2003; Granata et al., 2002). Greater knee energy absorption in females have been interpreted to in part explain the sex discrepancy in ACL injuries (Decker et al., 2003; Devita et al., 1992). Regardless of whether these sex differences can be attributed to cause lower extremity injuries in females, it is suggested that males and females utilize different landing strategies to attenuate the ground reaction forces at impact.

The head, arms, and trunk segment (HAT) is commonly overlooked as a portion of the kinetic chain within lower extremity injury research. The HAT segment generally comprises approximately 60% or greater of the total body mass (LeVeau, 1992), yet it is rarely accounted for in lower extremity injury research. Examination of factors controlling the HAT segment may therefore lead to improvements in controlling the body's center of mass and result in enhanced postural control. In addition, the position and orientation of all segments in the human kinetic chain influence the vertical ground reaction forces in landing (McNitt-Gray, Hester, Mathiyakom, & Munkasy, 2001). So a closer examination of factors influencing the HAT is further supported as there is potential for the HAT segment to influence the lower extremity kinetics.

Core stability describes the control and coordination of the lumbopelvic and hip complex (Hodges, 2003). There are three interdependent levels of core stability that are co-dependent on one another: 1) local spinal control, 2) lumbopelvic control, and 3) postural control (Richardson, Hodges, & Hides, 2004). Dysfunction at any level (local spine or lumbopelvic) may affect the other levels throughout the kinetic chain and ultimately affect postural control or whole body equilibrium. Loss of postural control in turn may lead to the subject falling or tripping and thus increasing the chances of sustaining a lower extremity injury.

When we consider the abdominal muscles responsible for maintaining appropriate core stability, the rectus abdominis (RA), external oblique (EO), internal oblique (IO), and transversus abdominis (TrA) muscles can be divided into two groups, local and global (Bergmark, 1989). The TrA and IO have attachments to the lumbar spine via the thoracolumbar fascia and are referred to as the local abdominals. The local abdominals are deep relative to the RA and EO and contribute to spinal, lumbopelvic, and postural control by increasing intra-abdominal pressure and sacroiliac joint stiffness (Bergmark, 1989; Cresswell, 1993; Cresswell & Thorstensson, 1994c; Richardson et al., 2002). The RA and EO are the global abdominals which contribute to lumbopelvic and postural control as they transfer forces to and from the thoracic cage and pelvis and ultimately control the HAT segment relative to the base of support (Bergmark, 1989; Hodges, 2003; Richardson et al., 2004). These different abdominal influences on the lumbopelvic and hip complex allow for their contribution to all three levels of core stability.

Outside of the biomechanical influences of the abdominals, local abdominal activation has been demonstrated to precede both global activation and that of upper and/or lower extremity movement (Hodges & Richardson, 1997a; Hodges & Richardson, 1997b). When the local abdominals are dysfunctional, particularly the TrA, global abdominal activations are delayed and their amplitudes increased (Ferreira, Ferreira, & Hodges, 2004; Hodges & Richardson, 1996). Not unlike the lower extremity injury literature, sex differences reported in the abdominal literature demonstrate that females have increased global abdominal contributions compared to males (Granata, Orishimo, & Sanford, 2001). As local abdominal dysfunction triggers global abdominal compensations, the resulting ability to maintain appropriate core stability at all three levels may be compromised. This global abdominal compensation mechanism may disrupt overall postural control and thus influence the lower extremity biomechanics. Interventions aimed to restore and/or facilitate local abdominal function may therefore optimize kinetic chain function.

Abdominal hollowing is a clinical technique that focuses on increasing local abdominal recruitment with minimal global abdominal involvement (Richardson, Jull, Hodges, & Hides, 1999). While this technique has been used to treat low back pain patients without the excessive global activation that can be detrimental to low back pain

(McGill, 1998) its influence on the rest of the kinetic chain through enhanced spinal, lumbopelvic, and postural control has not been established.

Statement of the Problem

Although the 'position of no return' is commonly viewed as a position that often results in ACL tears, the contributions of the trunk to this injury position are rarely considered within the scientific community (Ireland, 1999; Ireland et al., 1997; McClay Davis & Ireland, 2003). Further, the effects of common core stabilization exercises on lower extremity biomechanics have not been examined. Abdominal hollowing is a clinically based core stabilization exercise prescribed to low back pain patients which augments transversus abdominis activation while enhancing spinal stiffness (Richardson & Jull, 1995; Richardson et al., 1999). This enhanced stiffness may also facilitate lumbopelvic and postural control resulting in safer and more efficient lower extremity function and ultimately reducing the chances of the trunk contributing to the "position of no return".

The purpose of this study was to investigate the effect of abdominal hollowing on lower extremity biomechanics between sexes during dynamic activity. Specifically the intention was to determine if increased activation of the transversus abdominis, the prominent muscle contributing to abdominal hollowing, would influence leg spring stiffness and the joint energetic demands imposed on the ankle, knee, and hip joints during a double leg landing task in males versus females.

Objectives

1. The first objective was to demonstrate that all males and females could perform abdominal hollowing appropriately during the double leg landing task.

   Hypothesis 1: Subjects will exhibit similar EMG muscle activation patterns with AH in standing (static) and the preactivation occurring 150ms immediately prior to landing (dynamic) activities.

   Hypothesis 2: Subjects will demonstrate increased transversus abdominis-internal oblique activation ratios during the set of drop landings following clinical instruction (intervention).

   Hypothesis 3: Subjects will demonstrate the ability to hold the abdominal hollowing contraction during preactivation and after impact with the force plate.

2. The second objective was to quantify the changes in leg spring stiffness between control and abdominal hollowing (experimental) conditions as well as between sexes during a double leg landing task.

   Hypothesis 4: Females will initially have decreased leg spring stiffness in the control conditions compared to males. In the experimental condition (abdominal hollowing) females will increase leg spring stiffness, resulting in non-significant sex differences in leg spring stiffness.

3. The third objective was to quantify the changes in lower extremity energetics at the ankle, knee and hip between control and abdominal hollowing (experimental) conditions during the impact and stabilization phases of a double leg landing task as well as between sexes.

Hypothesis 5: Females will initially (control conditions) have increased energetic demands at the knee compared to males. With abdominal hollowing (experimental condition) females will have significantly reduced energetic demands at the knee resulting in non-significant sex differences.

Assumptions & Delimitations

1. Surface EMG will be used to monitor the electrical activity of the selected abdominal muscles.
2. The surface electrode placement for the Transversus Abdominis - Internal Oblique muscles represents electrical activity of the lower portions of both muscles and is representative of local abdominal activation.
3. Only the right leg and right abdominals will be studied. It is assumed that abdominal EMG and lower extremity biomechanical measures from the right and left sides will be symmetrical.
4. Joint energy absorption reflects both active (muscle) and passive (bone, articular cartilage, and ligaments) components.
5. Each of the abdominal muscles will be normalized to a percentage of its submaximal contraction and then represented as a proportion of the summed total normalized abdominal muscles.
6. Only recreationally active (exercise 30 minutes at least 3 times per week) subjects with present or past experience in jumping and landing sports and/or activites will be studied. The subject age range will be limited to 18-34.

Limitations

1. Abdominal activation ratios are interpreted as relative abdominal contributions and interpretations of absolute amplitude contributions from each muscle cannot be made.
2. Results from this dissertation cannot be generalized to populations other than recreationally active, college aged individuals.
3. Results cannot be generalized to activities other than the drop landing.
4. Interpretations of joint energetics cannot be attributed solely to active or passive components.

Operational Definitions

<u>Core</u> – The collective anatomical group encompassing the lumbar spine, pelvis, and hip joints and all the associated musculature and ligamentous components.

<u>Transversus Abdominis - Internal Oblique</u> <u>(TrA-IO)</u> – Local abdominal muscles represented by surface electromyography located 2 centimeters medial to and inferior to the right anterior superior iliac spine and superior to the inguinal ligament (Marshall & Murphy, 2003; McGill, Juker, & Kropf, 1996; O'Sullivan, Twomey, & Allison, 1998).

<u>Rectus Abdominis (RA)</u> – Global abdominal muscle represented by surface electromyography located at 2 centimeters lateral to the umbilicus (Dankaerts, O'Sullivan, Burnett, Straker, & Danneels, 2004; McGill et al., 1996).

<u>External Oblique (EO)</u> – Global abdominal muscle represented by surface electromyography located at 12 centimeters lateral to the umbilicus (Marshall et al., 2003; McGill et al., 1996).

Landing Impulse - The time period from initial contact with the force plate (exceeding 40 newtons of force) to when the body's center of mass reaches its lowest vertical position relative to the force plate (Zhang, Bates, & Dufek, 2000).

Impact Phase – The first 100 milliseconds of the landing impulse, (initial contact -> 100ms) (Decker, Torry, Noonan, Riviere, & Sterett, 2002; Decker et al., 2003). Unit = seconds.

Stabilization Phase – The time period from the end of the impact phase to when the body's center of mass reaches its lowest vertical position relative to the force plate (Kulas et al., *in preparation (a)*). Unit = seconds.

Leg Spring Stiffness (LSS) – The proportion of the peak ground reaction force divided by the body's center of mass displacement during a landing task (Farley & Morgenroth, 1999) and normalized to weight. Unit = Newton / meters.

Joint Powers (Hip, knee, and ankle) – The product of the internal moment and angular velocity at a joint (Devita et al., 1992; Winter, 1990). Unit = (Newton * Meter) / Second or Watts.

Energy Absorption (Hip, Knee, and Ankle) – The area under a joint power curve during the landing impulse (Devita et al., 1992; Zhang et al., 2000). Unit = Newton*meter.

Kinetic Chain – A conceptual framework that describes the influences and interactions between the ankle, knee, hip, pelvis, trunk and all other body segments.

CHAPTER II

REVIEW OF LITERATURE

Introduction

The goal of this dissertation is to provide an integrated link between core stabilization and lower extremity function. Therefore, this project will quantify the effect of Abdominal Hollowing (AH) on lower extremity biomechanics using a double leg landing model. This review of literature will provide: 1) a background of core stabilization from its development to its current state of clinical relevance, 2) a review of common lower extremity biomechanics utilized to explain injury potential in double leg landings, and 3) a background of scientific evidence to support the kinetic chain approach to understanding human movement leading to a rationale for the potential influence of AH on lower extremity biomechanics.

Core Stabilization

Introduction

Core stabilization refers to the muscles that act to stabilize the lumbar spine and lumbopelvic and hip complex as well as muscles acting to control position of the head, arms, and trunk (HAT) segment relative to the body's base of support (Hodges, 2003). Thus, core stability consists of three components: 1) lumbar spinal stabilization, 2) lumbopelvic control, and 3) postural control (Richardson et al., 2004). Although there are many contributing aspects to core stability such as lumbar musculature, and active and passive components of the spine itself, this discussion focuses mainly on the abdominal contributions to core stabilization.

Global versus Local Abdominal Function

The abdominal muscles are grouped according to their mechanical influences on the torso and spine. The global abdominal system is comprised of the rectus abdominis (RA), external obliques (EO), and internal obliques (IO), while the local abdominal muscle is primarily the transversus abdominis (TrA) (Bergmark, 1989). The RA and EO have extensive attachments from the thoracic cage to the pelvis and therefore function to provide a flexor moment on the trunk (Bergmark, 1989). Due to the structure and orientation of the global abdominals muscles, it has been suggested that they contribute to the transfer of forces to and from the thoracic cage and pelvis (Bergmark, 1989). This function of the global abdominals will be discussed later when examining a kinetic chain approach to understanding human movement. A second function of the global abdominals involves postural control of the head, arms, and trunk segments (HAT) relative to the pelvis and lower extremities (Hodges, 2003; Hodges & Richardson, 1999b).

Evidence of a global abdominal strategy of maintaining overall postural control has been demonstrated in studies involving upper and lower limb movements and the associated postural responses of the abdominals (Hodges, Cresswell, & Thorstensson, 1999a; Hodges et al., 1997a; Hodges et al., 1997b). These studies involved reaction-time tasks where subjects were first given a visual warning stimulus followed by a visual movement stimulus. When the movement stimulus was initiated, subjects were required to move a straight arm (Hodges et al., 1997b)or leg (Hodges et al., 1997a) as fast as possible in the direction indicated by the stimulus. Examination of the temporal patterns of the global muscles (RA, EO, and IO) demonstrated that although these muscles activated prior to initiation of the prime mover (indicated by deltoid or hip flexor onset), their onset and amplitudes were dependent on the direction of movement. Displacement of the HAT segment was not measured in these studies, but it was suggested that the global abdominals activate to offset the perturbations of the body's center of gravity caused by the prime mover (Hodges et al., 1997a; Hodges et al., 1997b).

In a follow up study, subjects were required to perform successive rapid upper limb movements while measurement of trunk and limb kinematics and EMG recordings were made (Hodges et al., 1999a). Results showed preparatory EMG activation of the global abdominals and erector spinae occurred according to the direction opposite of arm movement. These EMG activations were followed by trunk movement in the opposite direction to that of arm movement demonstrating the role of the global abdominals in controlling trunk movement (Hodges et al., 1999a) and thus supporting the theory that the globals activated to offset the perturbations in the body's center of gravity.

In all these studies assessing the abdominal postural responses to limb movement, TrA activation preceded that of all global abdominals and prime movers and its amplitude was invariant to the direction of movement (Hodges et al., 1999a; Hodges et al., 1997b; Hodges et al., 1997a). Therefore it was suggested that the local abdominal

(TrA) served a different and independent function from the globals by directly contributing to lumbar spine stability (Hodges et al., 1999b).

Although the global abdominals can contribute to core stability through lumbopelvic and postural control, the local abdominal system has the ability to contribute to core stabilization at all three levels: 1) local spinal control, 2) lumbopelvic control and 3) postural control (Richardson et al., 2004). However, the mechanisms of the local abdominal to contribute to core stabilization differ from that of the global abdominals. Because the local abdominal system contributes to all three components of core stability, a focused review of Transversus Abdominis' function and its clinical applications will now be presented.

Transversus Abdominis Function

The functional importance of the Transversus Abdominis (TrA) is demonstrated in multiple areas of research ranging from its role in respiration (DeTroyer, Estenne, Ninane, Van Gansbeke, & Gorini, 1990; Strohl, Mean, Banzett, Loring, & Kosch, 1981), to the development of intra-abdominal pressure (Cresswell, 1993), to maintaining sacroiliac joint stiffness (Richardson et al 2002)(Hodges et al., 1999a). This review will provide an overview of the functions of this muscle and how they relate to the three levels of core stability as proposed by Richardson et al 2004.

Respiratory Contributions

The earliest work of TrA function assessed its role in respiration. Both intramuscular and surface EMG have been used to evaluate the contributions of the abdominal muscles (RA,EO,IO, and TrA) to respiration during voluntary and involuntary contractions (DeTroyer et al., 1990; Strohl et al., 1981). These studies used real-time ultrasound to visually confirm proper intramuscular EMG placement. Results indicated that the TrA amplitudes were higher during expiration, hyperoxic hypercapnia (producing involuntary expirations), and against an inspiratory elastic load relative to the RA, EO, and IO muscles that coincided with a decrease in intra-abdominal pressure (DeTroyer et al., 1990). In addition, during the hyperoxic hypercapnia and inspiratory elastic loading conditions, there was a marked decrease in abdominal circumference as monitored qualitatively by the examiner. These studies were the first to confirm that the TrA is recruited differently from the global abdominal muscles during breathing. However, the mechanism by which the TrA contributed to intra-abdominal pressure and its functional importance was uncertain.

Development and Functions of Intra-Abdominal Pressure (IAP)

Activation of the TrA increases intra-abdominal pressure (IAP) during dynamic trunk loading and lifting tasks (Cresswell, 1993; Cresswell et al., 1994c). These studies assessed IAP intra-gastrically with a pressure transducer while muscle activation patterns of the RA, EO, IO, TrA, and erector spinae were monitored through intra-muscular EMG. Voluntary trunk extension efforts increased IAP with concomitant activity of the TrA and Internal Oblique (IO) whereas all the abdominals contributed to increasing IAP during the flexion effort (Cresswell, 1993). During a dynamic task similar to a straight leg dead lift, IAP was significantly positively correlated with TrA activity over different lifting and lowering velocities (r=.970). A similar relationship was found between IAP and IO, (r=.949) whereas the EO/IAP relationship was moderate (.637) and RA did not correlate with IAP (-.045) (Cresswell et al., 1994c). Functional importance of IAP was demonstrated by the linear correlation (r=.899, P<.05) of IAP and force (as measured by a load transducer in the pulley cable) (Cresswell et al., 1994c). Supporting the biomechanical contribution of IAP to the development of extension force, in-vivo phrenic nerve stimulation with no or minimal involvement from the abdominal muscles results in increased IAP as well as a moderate extensor torque at L3 (Hodges, Cresswell, Daggfeldt, & Thorstensson, 2001). Although, it was reported that the global abdominals (EO) can contribute to the increases in IAP (Cresswell et al., 1994c), excessive co-contraction of the globals may in fact cause the lumbar spine to become unstable and/or "buckle" due to excessive compressive forces on the lumbar spine (McGill, 1998). Therefore, as IAP generation is important functionally during lifting and lower movements, it may be suggested that this should be done through focused TrA activation rather than global abdominal activation in order to protect the lumbar spine from injury.

Increases in IAP have been demonstrated to be beneficial not only in force development during trunk extension efforts (Cresswell et al., 1994c), but also in maintaining postural control (Cresswell, Oddsson, & Thorstensson, 1994b; Hodges, Cresswell, & Thorstensson, 2004). During expected and unexpected tasks, Cresswell et al (1994) monitored muscle activity, IAP, and trunk displacement when the trunk was perturbed through a system of cables attaching ventrally and dorsally to the chest via a vest (Cresswell et al., 1994b). During self-initiated loading conditions, the abdominals preactivated prior to trunk movement which coincided with an IAP increase and a smaller angular trunk displacement compared to the unexpected conditions. During the unexpected conditions, IAP development was delayed with subsequent larger angular trunk displacements. Although later in activation onset relative to the unexpected perturbations, it was demonstrated that the TrA and Obliques contributed to IAP development and thus controlling trunk displacement.

Supporting the connection of IAP development to control of trunk motion, Hodges et al (2004) used a support-surface perturbation where the floor would move in anterior, posterior, medial, and lateral directions and found that IAP increases were directional and amplitude dependent suggesting a IAP response mechanism to provide postural stability (Hodges et al., 2004). Although abdominal EMG was not utilized in this study, the TrA and IO were suggested to contribute to this postural response as they have been most closely associated with IAP development (Cresswell, 1993; Cresswell et al., 1994c).

The TrA increases IAP through its contraction and subsequent tensioning of the thoracolumbar fascia (Hodges, 2003). Due to transverse fiber orientation, the TrA acts as a corset and effectively decreases lower abdominal circumference while increasing IAP (DeTroyer et al., 1990; Richardson et al., 2004). This corset-like action is referred to as the abdominal hoop mechanism (Figure 1) (McGill, 2002; Richardson et al., 1999). The transverse orientations of the TrA and lower portions of the IO have been demonstrated through cadaver morphology (Urquhart, Barker, Hodges, Story, & Briggs, 2005). In addition, TrA and IO attachments to the lumbar spine via the thoracolumbar fascia were also observed (Urquhart et al., 2005) thus supporting the mechanical effects of the TrA on the modulation of IAP. Also because of the structure and orientation of the IO, some have considered the lower fibers of this muscle as a local abdominal along with the TrA (Bergmark, 1989; Marshall et al., 2003)

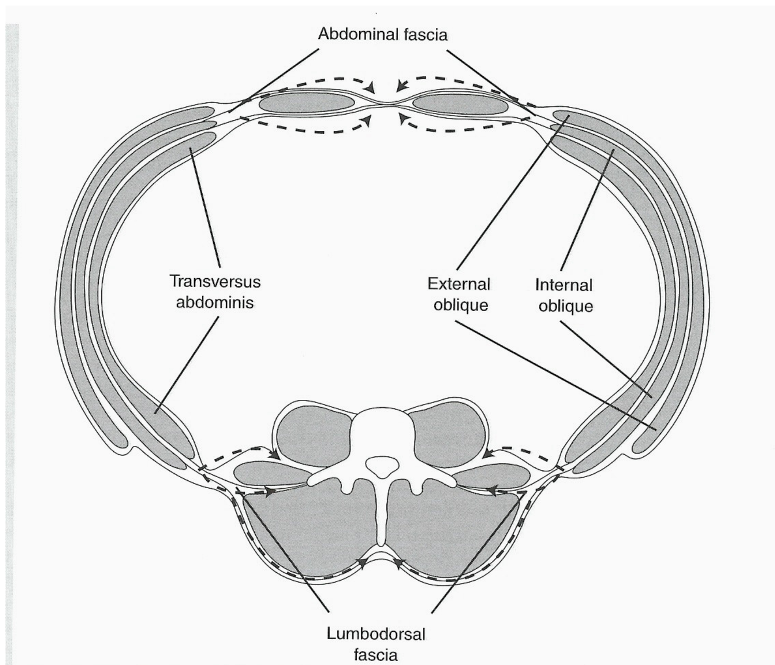

Figure 1 The Abdominal Hoop Mechanism (adapted from McGill, 2002)

Because of its attachments on the lumbar spine via the thoracolumbar fascia and its role in IAP development, TrA activation has been demonstrated in vivo to increase lumbar spinal stiffness directly (Cholewicki & McGill, 1996; Hodges et al., 2003; Hodges et al., 2001). In a porcine model, stimulating electrodes were directly attached to the TrA while pins were fixated to L3 and L4 with motion sensors attached on the distal aspect of the pin to monitor lumbar displacement (Hodges et al., 2003). To adequately measure stiffness at L3/L4, the pins were fixated to a motor to evoke lumbar movement. Direct stimulation of the TrA and diaphragm (on separate trials) resulted in increased IAP (as measured by a pressure catheter) such that the motion between L3 and L4 was decreased when comparing to control conditions with no electrical stimulation. These results demonstrate that IAP development via diaphragm or TrA stimulation increases local spinal control directly.

Sacroiliac Joint Stiffness

TrA activation has been demonstrated to increase sacroiliac joint stiffness (Richardson et al., 2002). This study measured the differences in vibration between the ilium and the sacrum to determine the stiffness of the joint. This measurement was based on the premise that if the stiffness of the joint was high, vibrations applied to the anterior superior iliac spine at 200hz would transmit across the sacroiliac joint and the two bones would vibrate at the same frequencies. A large difference in frequencies between the ilium and sacrum would therefore indicate a less stiff joint. Differences in stiffness measures from relaxed condition to TrA activation condition demonstrated that the sacroiliac joint was indeed stiffer in the TrA condition (Richardson et al., 2002).

Building on its role in intervertebral stiffness generation (Hodges et al., 2003) and sacroiliac joint stiffness (Richardson et al 2002), the TrA seemingly also has the ability to control lumbopelvic motion. As the pelvic floor forms the lower border of intra-abdominal cavity, any increases in IAP may also affect pelvic motion relative to the lumbar spine. In addition, the lower fibers of the TrA have attachments on the iliac crest suggesting a direct biomechanical influence on the pelvis when activated (Urquhart et al., 2005). Therefore, the motion between the pelvis and lumbar spine may be closely associated in subjects with a properly functioning TrA.

In the assessment of the relationship between the static lumbar spine lordosis and pelvic inclination, Gardocki et al (2002) assessed 36 sagittal plane radiographs and demonstrated that total segmental lumbar lordosis significantly correlated (r=.82) with lumbopelvic lordosis (a combined measure of pelvic inclination and lumbar lordosis) (Gardocki, Watkins, & Williams, 2002). Although these results support a close relationship between lumbar lordosis and pelvic inclination, the largest limitation with this study was that this static lumbo-pelvic relationship may not necessarily exist in dynamic motion. To date there are no known studies closely evaluating lumbar spine motion as it specifically relates to sagittal pelvic motion (McGill, 2002). However, the collective functions of the TrA to increase IAP, intervertebral stiffness, and sacroiliac joint stiffness suggests that the TrA contributes to lumbo-pelvic control.

<u>Summary</u>

This review of TrA function demonstrated its biomechanical influence in all three aspects of core stabilization: 1) local spinal control (Cresswell et al., 1994b; Hodges et al., 2003), 2) lumbopelvic control (Richardson et al., 2002), and 3) postural control (Cresswell et al., 1994b; Hodges et al., 2004). The mechanism by which the TrA accomplishes this is through the abdominal hoop mechanism effectively increasing IAP.

This discussion highlights the need to explore clinical postures and/or exercises that not only focus on the global muscle system, but also the re-education and facilitation of the local abdominal system. A clinical postural exercise aimed at re-educating the local abdominal system is abdominal hollowing and will be discussed in the next section.

Abdominal Hollowing

Abdominal hollowing (AH) is a clinically based exercise that facilitates TrA activation in isolation relative to the other global abdominal muscles (Richardson et al., 1999; Strohl et al., 1981). This exercise aims to restore the normal function of the TrA and has had clinical relevance in patients treated for low back pain (McGill, 2001). Over the past 15 years, developments of this exercise and its clinical implications have changed dramatically. The earliest reports evaluated muscular contributions to abdominal hollowing, using fine-wire intramuscular EMG with real-time ultrasound confirmation, demonstrating that this clinical exercise places an emphasis on the transversus abdominis (DeTroyer et al., 1990; Strohl et al., 1981). Because of the obvious difficulty and clinical impracticality of using intramuscular EMG to observe abdominal function, research has evolved to better assess the contribution of the TrA to enhance spinal stiffness and thus lumbar stabilization through real-time ultrasound (Richardson et al., 1995). Contrary to prior thought that the IO contributed primarily to the global abdominal system (Bergmark, 1989), it is currently thought that the lower portions of the IO also have a local stabilization component because of the attachment to the lumbar spine via the thoracolumbar fascia (Richardson, Toppenberg, & Jull, 1995) and horizontal orientation of these fibers below the level of the anterior superior iliac spine (Urquhart et al., 2005). Additionally, two studies examining the morphology of the abdominals in cadavers reported that the lower fibers of the TrA and IO were fused in several specimens (Marshall et al., 2003; Urquhart et al., 2005). For these reasons the remainder of this discussion will refer to both the TrA and IO as contributors to the local abdominal system. This section will focus on the development of AH as a means to clinically evaluate local abdominal function.

Clinical Evaluation of Abdominal Hollowing

In order to differentiate local abdominal recruitment from the global recruitment, Richardson & Jull's research team utilized a pressure biofeedback unit (The Stabilizer, Chattanooga, TN) to isolate local abdominal activation in a supine position (Richardson, Jull, Toppenberg, & Comerford, 1995). Using this method of assessing lumbar stabilization, the clinician placed the biofeedback unit under the lumbar spine of a supine patient and inflated the unit so that the bag fit snuggly under the lumbar spine (Figure 2).

Figure 2 Abdominal Hollowing in Supine to Monitor Lumbar Pressure

Lumbar stabilization was then assessed as the patient performed various leg lifting motions, with simultaneous performance of the AH maneuver. No increase in pressure on the bag was the desired clinical outcome. An increase in pressure qualitatively indicated rectus abdominis recruitment (global muscle substitution). Although the supine position allowed the clinician to focus on the oblique abdominals and transversus abdominis, this position was not sensitive enough to ensure the TrA and IO were integral to local abdominal stabilization.

In order to better focus on proper AH activation patterns, the pressure biofeedback unit was placed under the lower abdomen with the patient in a prone position (Figure 3) (Richardson et al., 1995). In order to precisely isolate the local abdominal system (TrA and IO), the patient was instructed to hollow or “draw in” the abdomen by pulling the navel up and in towards the spine (Richardson et al., 1999). A decrease in pressure would ensure that the TrA and IO were activated preferentially to the rectus abdominis. In support of this technique, researchers confirmed abdominal hollowing (AH) simultaneously with real-time ultrasound and showed a thickening of the TrA with some IO thickening; and minimal EO and RA thickness changes (Richardson et al., 1999). Others have also recently reported that the RA and EO are minimally recruited during the AH maneuver supporting a focus on local abdominal involvement which suggests the local abdominals (TrA & IO) are the primary muscles used in AH (Drysdale, Earl, & Hertel, 2004).

Figure 3 Prone Quantification of AH

The primary activation of the TrA and IO during AH has been confirmed with real-time ultrasound in four point kneeling, (Beith et al., 2001; Critchley, 2002) supine, and prone positions (Richardson et al., 2002; Richardson et al., 1999). These studies collectively concluded that four-point kneeling is a reliable and valid means of instructing AH to patients naïve to the activation pattern (Beith, Synnott, & Newman, 2001; Richardson et al., 1999). Most recent reports indicate a linear relationship between visually observing the thickening of the TrA through real-time ultrasound and intramuscular EMG of the TrA (McMeeken, Beith, Newham, Milligan, & Critchley, 2004).

Since the development of using real-time ultrasound and intramuscular EMG to measure TrA activation, Stuart McGill and colleagues examined whether surface EMG could represent the muscle activation patterns associated with the TrA (McGill et al., 1996). This involved comparisons of TrA intramuscular EMG to IO surface EMG. The surface location of the IO was located just superior to the inguinal ligament. Activation profiles of the two muscles were compared during maximum voluntary contractions (MVC). Results showed that the intramuscular TrA and surface IO had root mean square (RMS) differences of up to 15% of MVC. Results from this study indicate that although there are differences in these two measures, researchers may use surface IA EMG to represent TrA activation as long as the 10-15% RMS differences between MVCs are acknowledged.

Since McGill et al (1996) reported the activation profiles of surface EMG of the IO and intramuscular EMG of the TrA, two other studies have tested the same surface location with the AH performance (Marshall et al., 2003; O'Sullivan et al., 1998). While the latter study conservatively reported this location (2cm lateral and inferior to the ASIS) to be representative of IO activation, the former reported this location as representing both TrA & IO activation. Both studies also assessed muscle activation of the RA and EO while performing the AH maneuver. Marshall & Murphy (2003) demonstrated that the location used to measure TrA & IO surface EMG can reliably replicate the feedforward response of the TrA reported elsewhere (Hodges, 2003). In addition, the TrA-IO surface electrode placement was reported as valid because of its higher amplitude relative to the RA and EO during AH (Marshall et al., 2003). The strength of this latter study though comes from the argument that this electrode placement of the IO is representative of both TrA and IO activation because the fibers of these muscles are fused at this location (Figure 1). Instead of attempting to differentiate these muscles, the authors reported it as a collective TrA – IO (Marshall & Murphy, 2003). This is supported in an earlier study, that reported the IO (same electrode placement as Marshall & Murphy 2003) as being the primary contributor with AH (O'Sullivan et al., 1998).

Both research groups described above report the use of surface IO as representing local abdominal activation during AH. As mentioned earlier, the IO attaches to the thoracolumbar fascia with the TrA (Richardson et al., 1999). In addition, the TrA and IO were demonstrated to be fused at the IO surface EMG location in 9 out of 10 cadavers (Marshall et al., 2003). The latest cadaver study assessing abdominal morphology did report this TrA-IO fusion although not occurring at the same frequency as Marshall &

Murphy (2003). However, the fascicle length reported in this study indicated that it extended 3.6cm medially from the anterior superior iliac spine (ASIS) before divisions of the IO started (Urquhart et al., 2005). For these reasons it is logical that the location used for surface EMG of the IO, 2cm medial and inferior to the ASIS and superior to the inguinal ligament, is representative of both TrA & IO activation. Since both muscles represent the local abdominal system, this surface location is appropriate to confirm the muscle activation associated with AH.

In summary, performance of the AH involves a relatively isolated activation of the local abdominal system. Early research has mentioned the TrA as the primary muscle contributing to this clinically prescribed exercise. However, some of the latest research suggests that the IO has a role in the local abdominal system as well. While this section focused on the development of AH as a clinical measure of local abdominal activation (TrA-IO), there is potential for practical applications as well.

Practical Applications for Abdominal Hollowing

AH has been used extensively within the clinical setting for treatment of populations with low back pain (McGill, 2002; Richardson et al., 1999). Clinicians have used AH as an assessment tool for appropriate local abdominal function and for re-education of patients with dysfunctional local abdominal activation patterns (Richardson et al., 1995). Efficacy of AH to target and restore local abdominal function has been demonstrated in low back pain populations (McGill, 1998; O'Sullivan, 2000; O'Sullivan et al., 1998). Methods by which AH and local abdominal activation are assessed have become more practical in the clinic with use of real-time ultrasound, pressure biofeedback units, and surface EMG (Richardson et al., 1999). In light of these clinical applications, AH and the local abdominal system may have applications in other areas as well.

Historically in the sporting world core stabilization regimens have incorporated exercises aimed at strengthening the abdominals (Cissik, 2002). However, most of these exercises consist of abdominal crunches and oblique twists with varying weights such as medicine balls and isotonic machines concentrating mainly on training the global abdominal system's total capacity (Hodges, 2003). Within collegiate strength and conditioning and athletic training communities local abdominal assessment and re-education is only recently becoming more understood (Johnson, 2002). As reports suggest that low back pain incidences may be higher in females as compared to males (Nadler et al., 2002a), the need exists for athletic trainers and clinicians to more closely observe local abdominal function in athletes. As others have demonstrated that TrA dysfunction has also been observed in patients without complaints of low back pain, this population may be "at risk" for future low back injury (McGill, 2002; Richardson et al., 1999). This focus on local abdominal function may be central in treating low back pain patients (Nadler et al., 2002a; Nadler, Wu, Galski, & Feinberg, 1998). Furthermore, abdominal exercises aimed at strengthening global abdominal capacity (i.e. crunches, weighted situps, etc) increase compressive loads to the lumbar spine which may be detrimental in populations with a dysfunctional local abdominal system (McGill, 1998; McGill, 2001). For these reasons, the AH may be useful as: 1) an assessment tool for asymptomatic athletes during pre-participation examinations, and 2) for re-education of athletes suffering from chronic low back pain exhibiting TrA-IO dysfunction.

Although practical applications of AH are numerous in the treatment of low back pain, there is little empirical evidence showing how the implementation of these core stabilization exercises affect the body's kinetic chain. To date a single pilot study suggests that AH has the potential to influence the lower extremity biomechanics (Kulas, Windley, & Schmitz, 2005). Large effect sizes were noted between subjects in AH and control groups for leg spring stiffness and knee energy absorption during a single leg landing task. Due to lack of power most likely due to a small sample size, we did not achieve statistical significance. Whether positive or negative, AH's potential influence on the lower extremity biomechanics may be feasible as enhanced spinal stiffness and lumbopelvic control may contribute to the control of the HAT segment relative to the lower extremity and therefore overall postural control. Prior to a discussion on the relationships between core stabilization and lower extremity function, it is necessary to review the current state of the landing literature and the associated lower extremity biomechanics.

Lower Extremity Biomechanics in Double Leg Landings

Introduction

Landing tasks require the body's musculoskeletal system to absorb the ground reaction forces (GRFs) at impact while terminating the body's vertical momentum (Dufek & Bates, 1990; Lees, 1981; Schot et al., 1993). While joint energetics are representative measures of how the hip, knee, and ankle musculature contribute to attenuate these GRFs, leg spring stiffness (LSS) has been thought of as a determinant of the GRFs (Farley et al., 1999; Zhang et al., 2000). These biomechanical variables have often been utilized in an effort to explain a person's injury potential in landing tasks (Butler, Crowell, & McClay Davis, 2003; Devita et al., 1992). The following sections will discuss: 1) a brief review of inverse dynamic calculations, 2) lower extremity energetics 3) leg spring stiffness, and 4) the inter-relationships between LSS and joint energetics as they potentially explain injury potential in landing tasks.

Inverse Dynamic Calculations

During activities that require the foot coming in contact with the floor (closed-kinetic chain), estimates of the individual joint forces and moments are calculated through an inverse dynamics solution (Winter, 1990). Three pieces of information are required to appropriately calculate the joint moments: 1) force data, 2) anthropometric data, and 3) position data. Force data are usually acquired through the use of a force plate. Anthropometric data are charted information that estimates segment mass, length, and radius of gyration (LeVeau, 1992). Each body segment, i.e. foot, tibia, femur, etc. have unique masses and lengths that are based on a person's true height and weight.

Position data refers to the positions of the segments within the testing environment. These data are traditionally acquired through the use of video analysis in which the positions of segment markers are tracked over time (Winter, 1990). More recently, electromagnetic tracking systems have been utilized to acquire position data as individual sensors are attached to bony segments such as the foot, tibia, femur, and sacrum (Madigan & Pidcoe, 2003).

Position data of the human segments are made possible by establishing two coordinate systems: global and local. The global or fixed coordinate system is defined by an orthogonal (X,Y,Z) axis system and provides the 3-dimensional environment that the human movement occurs within. Using electromagnetic measurement equipment, each individual sensor represents a rigid body segment. A local coordinate system for each body segment is used to establish the segment's location (Z, Y, X) and orientation (rotation around each Z, Y, and X axis) within the global coordinate system's environment (Allard et al., 1995).

Once the force, anthropometric, and position data have been acquired the joint moments can be calculated which represent the internal moment required by the active (muscle) and passive (ligament, capsular) structures of the given joint to overcome the ground reaction and external forces imposed on the joint (Winter, 1990). Based on these calculations of internal joint moments, calculation of the joint powers (internal moment multiplied by angular velocity of the joint) and joint energetics (power multiplied by time) are possible. These variables and their calculations have been used as a measure of how the body's musculoskeletal system absorbs the impact forces at landing (Schot et al., 1993).

Lower Extremity Joint Energetics

The negative mechanical work performed on the hip, knee, and ankle joints reflect the ability of the active and passive restraints to absorb the impact forces at landing (Schot et al., 1993). The hip extensors, knee extensors, and ankle plantar flexors are the primary muscles contributing to this energy absorption during landings (McNitt-Gray et al., 2001). Through energy absorption, the importance of these lower extremity muscles are highlighted as they eccentrically control end ranges of joint motion by decelerating the proximal and distal segments (Prilutsky, 2000). Thus, as individual joint powers indicate which muscles and joints are primarily contributing to the negative mechanical work, these variables have been utilized in research focused on injury potential (Decker et al., 2003).

Examination of the literature involving hip, knee, and ankle joint energetics during landings reveals mixed results. The knee has often been shown to be the most consistent energy absorber in landing tasks (Minetti, Ardigo, Susta, & Cotelli, 1998; Zhang et al., 2000). However, others have noted that the ankle is the primary energy absorber in "stiff" landings and landings from higher heights while the hip contributes more relative energy absorption during landings characterized as "soft" (Devita et al., 1992; Zhang et al., 2000).

Sex differences have also recently been reported within the lower extremity energetic literature (Decker et al., 2003). The major findings of this study indicated that females primarily utilize the knee and ankle to absorb energy at landing whereas males had no significant absorption differences by joint. To date, this is the first and only study that was found comparing males and female landing differences using energetic variables. The differences in energy absorption in this study were purported to possibly explain the higher knee injury incidences in females commonly reported in the literature (Griffin et al., 2000). Other studies utilizing energy absorption have generally used either males or females as the subject population, and comparisons between these studies are difficult. Others have reported that females consistently utilize the knee as the primary energy absorber while the ankle is the secondary energy absorber (Kulas et al., in preparation (a)). While Devita & Skelly reported that the ankle is the primary energy absorber in females (Devita et al., 1992), the results of this study are confounded by methodological differences as the subjects wore sneakers during the landings which may have influenced the joint energetics.

Emphasizing the high energy absorption at the knee and ankle, studies utilizing a female subject population report that the hip contributes only about 7-25% of the total joint energy absorption (Decker et al., 2003)(Kulas et al., in preparation (a)). Conversely, hip energy absorption in males ranges from 30-40% of the total joint energy absorption (Decker et al., 2003; Zhang et al., 2000). Together these studies emphasize that while males seem to distribute the GRFs at impact more evenly among the ankle, knee, and hip, females seem to primarily utilize the knee and ankle joint to absorb the forces at impact. These differences in energy absorption between sexes (Decker et al., 2003) have been hypothesized to be a reason why females sustain more ligamentous knee injuries relative to males (Arendt, Agel, & Dick, 1999).

A major caution should be noted in comparing studies involving energetics because the energy absorption at landing is dependent on the body's total kinetic energy prior to landing which can be manipulated by the landing height (McNitt-Gray, 1993; Zhang et al., 2000) and length of time used to calculate energy absorption (Winter, 1990). While most landing studies utilize 60cm as a landing height (Decker et al., 2002; Decker et al., 2003; Zhang et al., 2000), length of time utilized to calculate the joint energetics range from the first 100ms of landing to when the body's center of mass position reaches its lowest point (171-260ms) (Decker et al., 2003; Devita et al., 1992; Zhang et al., 2000)(Kulas et al., in preparation(a)). To remedy these differences in landing times, researchers have either normalized the energetics to a percentage of total landing phase (Zhang et al., 2000) or computed an average joint power based on the landing phase times (Kulas et al., in preparation (a)).

In addition to examining how the impact forces are distributed between the joints at impact, the sequence of energy transfer during landing occurs from the distal ankle to the proximal hip (Prilutsky & Zatsiorsky, 1994). This energy transfer is made possible through the biarticular gastrocnemius and rectus femoris muscles (Prilutsky et al., 1994; Winter, 1990). This transfer of energy is supported in studies utilizing stiff and soft landings. Stiff landings are usually dominated by energy absorbed at the ankle and knee while soft landings utilize more hip and knee absorption (Devita et al., 1992; Zhang et al., 2000). In landing tasks characterized as soft, the hip undergoes greater hip flexion placing an increased stretch on the hip extensors and giving them a mechanical advantage to better contribute to energy absorption. Conversely, stiff landings are characterized by relatively less hip flexion and more knee flexion and ankle plantar flexion thus increasing the energy absorption demands on the gastrocnemius and quadriceps muscles.

Individual joint energetics enable the researcher to assess which joint preferentially absorbs the GRFs at landing. Landing models have often been implemented that utilize stiff and soft landing techniques from same heights (Devita et al., 1992; Zhang et al., 2000)(Kulas et al., in preparation (a)). Through a within subject model in which the height is fixed so the kinetic energy at landing is constant across trials, changes in joint energetics can be attributable to the active musculoskeletal and passive ligamentous and capsular restraints of each joint (Devita et al., 1992). Finally, the rate of energy absorbed on these lower extremity muscles, appropriately termed mechanical power (Winter, 1990) reflects the magnitude of loading at each joint and thus may indicate the joint's potential for injury at landing. As this section reviewed the individual joint contributions to landing, this next section will review a whole-body measure of how the vertical momentum is decelerated in landing, termed leg spring stiffness.

Leg Spring Stiffness

Leg Spring Stiffness (LSS) is defined as the peak GRF divided by the body's center of mass vertical displacement (Farley & Ferris, 1998a). This variable has been used in hopping and landing models and reflects the ability of the active musculoskeletal system to decelerate the body's momentum at landing (Farley et al., 1998b; Farley et al., 1999; Granata et al., 2002). Because LSS has been thought to be a determinant of the potentially injurious GRFs at impact, researchers have postulated that LSS may be an important variable in terms of injury potential in landing (Butler et al., 2003; Farley et al., 1999). This section will review the available literature and examine its potential clinical relevance.

Injury potential in landing models has been thought to be related to the magnitude of GRFs in landing (James, Bates, & Dufek, 2003). LSS is directly affected by the GRFs at impact and quantifies the interaction between the peak GRFs and the body center of mass displacement in landing. There is speculation that high LSS values may place an individual at risk for bony injury while excessively low LSS values indicate increased potential for soft tissue injury (Butler et al., 2003). Although it has been theorized that an optimal LSS magnitude may be most beneficial to minimize risk for musculoskeletal injury, this optimal magnitude is not known (Butler et al., 2003).

LSS can be influenced through the manipulation of: 1) ground contact time in hopping (Arampatzis, Schade, Walsh, & Bruggemann, 2001; Granata et al., 2002) and 2) stiff & soft instructions in landing tasks (Kulas et al. in preparation (a)). These studies all utilized a preferred landing condition to serve as within subject controls. Results reported by Granata et al (2002) show that females hopped with less LSS than males (Granata et al., 2002). However, these sex differences were mostly attributed to anthropometrics and have been supported by others researching sex differences in stiffness (Blackburn, Riemann, Padua, & Guskiewicz, 2004). Nevertheless, the ability to change LSS within subjects may lend insight into a person's risk for injury. If there was a known relationship between individual joint energetics and LSS, a better interpretation of LSS may help to clarify what people or joints may be at risk for injury.

Inter-Relationships between Leg Spring Stiffness and Joint Energetics

Individual joint energetics and LSS have been implicated in the explanation of injury potential in recreationally active populations (Decker et al., 2003; Granata et al., 2002). While excessively high and low LSS magnitudes have been implicated to place an individual at risk for bony and soft tissue injuries respectively (Butler et al., 2003), it is unclear as to which lower extremity joints may be at risk.

In order to assess individual joint contributions comprising LSS, Kulas and colleagues used a within-subject design to assess changes in LSS and joint energetics across preferred, stiff, and soft landing techniques (Kulas et al., in preparation (a)). Findings of this study indicated that although LSS was different across preferred, stiff, and soft conditions, lower extremity joint average powers were not. In fact, the knee had the highest average powers and was the primary energy absorber regardless of landing technique during both the impact and stabilization phases of landing. A regression analysis revealed that while the knee explains 60% of the variance in LSS over both phases, hip absorption during the stabilization and ankle absorption during the impact phase combined to predict an additional 15% of the variance in LSS. By utilizing this regression model individual joint energy absorption predicted a total of 75% of the variance in LSS. This study was conducted using a female dance population so generalizations across genders should be avoided. It is of clinical importance that females consistently had the highest loading rates imposed on the knee regardless of landing technique. This consistent absorption strategy may partially place the knee at risk for injury. Therefore, efforts to more effectively distribute the absorption demands across the ankle, knee, and hip may decrease chances of injury at the knee joint. For future studies an intervention aimed to enhance the joint energetics at the hip or ankle seems to be a logical method of testing the ability to redistribute joint absorption.

The results of this study indicated that lower LSS results in higher energy absorption at the knee, hip, and ankle while higher LSS indicates decreased energy absorption at the joints. Because joint energetics indicate both active and passive musculoskeletal contributions in landings (Winter, 1990), high energy absorption may place a person at risk for soft tissue or ligamentous injury. Taking into account that this study followed a repeated measures design, subjects landed with approximately the same total kinetic energy across each of the three conditions as indicated by the fixed box height, low total energy absorption (and high LSS) and may be indicative that the GRFs are primarily attenuated by the bony and inert joint structures thereby placing someone at risk for bony injury.

The relationships between LSS and individual energetics are likely valuable in terms of assessing injury potential and allowing interpretation of which joints may be susceptible to injury. From an injury risk and/or prevention point of view, it may be advantageous to gain a better understanding of how to influence these variables so that an "optimal" pattern of joint absorption and thus whole body deceleration may occur to minimize injury potential.

This section outlined lower extremity contributions to dissipate the GRFs during landing. However, while the head, arms, and trunk (HAT) are also part of the multi-linked human body and comprise 66% of the total body's mass (LeVeau, 1992), the influence of this HAT segment on these lower extremity biomechanics have not been established. The third section of this review of literature will take a functional approach to assessing human movement by considering the influence of the HAT segment via core stabilization on lower extremity biomechanics.

A Kinetic Chain Approach to Understanding Human Movement

Introduction

The first two sections of this review of literature discussed: 1) core stabilization in regards to global and local abdominal function, and 2) lower extremity biomechanics as they relate to injury potential in landing tasks. This third section of the review will explore concepts that have been used to explain human movement as an integrated kinetic chain. Using a kinetic chain approach this review will conclude with how core stabilization can influence lower extremity biomechanics.

Biomechanical Performance Assessment

Researchers assessing the biomechanics of jumping and throwing tasks support a sequence of proximal to distal influences to achieve maximal performance. These influences support a dynamic kinetic chain approach to understanding human movement. Bobbert & Schenau (1988) evaluated jumping coordination to attain maximum vertical height and found that the hip, knee, and ankle's extension movement, moment, and muscle activation patterns occur in a proximal to distal sequence (hip, knee, ankle) (Bobbert & van Igen Schenau.G.J., 1988). To explain this kinetic chain assessment, the roles of the biarticular muscles i.e. rectus femoris, gastrocnemius, and hamstrings have been demonstrated to be responsible for the transfer of powers from one joint to an adjacent one in order to maximize take-off velocity of the body's center of mass (Bobbert et al., 1988; Jacobs, Bobbert, & van Igen Schenau, 1996). For example, in a leg extension invoked by the quadriceps muscle, the gastrocnemius is stretched posterior to the knee at its origin. In order for the gastrocnemius to maintain its same resting length (origin to insertion), the ankle plantar flexes. In this simplified example force was transferred from the knee to the ankle to achieve plantar flexion via the gastrocnemius although the subject did not actively plantar flex the ankle (Jacobs et al., 1996).

Contrary to jumping coordination patterns, landing coordination has been demonstrated to involve a distal to proximal transfer of mechanical energies to dissipate the vertical ground reaction forces (GRFs) at impact (Prilutsky et al., 1994). This transfer was demonstrated as the percentage of energy absorption calculated at the hip was effectively 39% of the total work done by the ankle joint showing that the proximal muscles and joints aid the distal ankle and knee in attenuating the GRFs at impact. From an injury risk perspective it may be advantageous to facilitate hip absorption so that the ankle and knee absorption demands are minimized.

Proximal to distal sequences of force development have also been demonstrated using biomechanical and neuromuscular analyses during throwing. In generation of forces large enough to pitch a baseball up to 90mph, it has been stated that the shoulder and upper extremity are incapable of developing such high accelerations alone, yet the shoulder joint has been recorded to reach a peak internal rotation velocity of 6100 degrees per second just prior to ball release (Feltner & Dapena, 1986). Therefore it has been suggested that the ground reaction forces developed in the lower extremity from pushing off the ground are transferred up the lower extremity through the torso via the global abdominals and into the upper extremity until the forces are ultimately transferred from the shoulder through the elbow and to the hand just prior to ball release (Bergmark, 1989; Feltner et al., 1986; Young, Herring, Press, & Casazza, 1996). To provide evidence of this, Alexander (1991) used a simple mathematical model to show that when trunk rotations are restricted, ball velocity is substantially decreased (Alexander, 1991) as opposed to that of others where ball velocities doubled in throwing motions involving the trunk (Atwater, 1979).

A proximal to distal force transfer has been demonstrated in pitching and others have subsequently applied this concept to shoulder rehabilitation (McMullen & Uhl, 2000). Rehabilitation exercises adopting this kinetic chain approach focus on using functional whole body motion to develop the forces in order to throw while the scapulothoracic and glenohumeral musculature stabilizes the shoulder (McMullen et al., 2000; Young et al., 1996). Proprioceptive neuromuscular facilitation techniques have also been developed which emphasize this proximal to distal muscle activation pattern (Adler, Beckers, & Buck, 1993). The mechanism that explains this proximal to distal approach is irradiation, which refers to the effect that a stimulus (such as a push or pull) has on other synergistic muscles (Adler et al., 1993). This concept as it is applied to rehabilitation techniques utilizes multi-jointed movements which promote kinetic chain function. In addition to the utilization of the proximal to distal approach for human movement performance, proximal muscle activations and movements preceding extremity movement have been demonstrated in the postural control literature.

Postural Control

Postural control describes the ability to maintain the body's center of mass (CoM) within its base of support (Latash, 1998). As the head, arms, and trunk (HAT) segments comprise approximately 66% of the total body's mass (LeVeau, 1992), its effect on postural control is apparent. This section will discuss the lower extremity contributions to postural control and finish with a discussion of how these influences are related to the abdominal postural responses supporting a kinetic chain approach to postural control.

<u>Lower extremity influences on postural control</u>

Postural synergies between the lower extremity and trunk exist in order to maintain the CoM within the base of support to ultimately maintain balance (Alexandrov, Frolov, & Massion, 2001; Latash, 1998). During normal stance, the body is often modeled as an inverted pendulum where the ankle musculature primarily controls the CoM position relative to the supporting feet (Aruin, Ota, & Latash, 2001). In this study, rotational and lateral perturbations to the trunk were created as subjects were required to: 1) catch/release a weight and 2) self-initiate high velocity arm flexion/extensions and horizontal abduction/adduction motions. Responses in the lower extremity showed that as modifications of soleus and tibialis anterior activations and deactivations were dependent on the direction and velocity of arm movement, an ankle strategy was demonstrated in order to maintain the body's CoM within the base of support and thus ensure adequate postural control (Aruin et al., 2001).

The hip and knee joints also have the ability to influence the position of the body's CoM. It has been reported that while ankle and hip strategies exist to maintain an upright posture in slipping-type perturbations, as evidenced through muscle activation patterns and production of ankle and hip torques (Runge, Shupert, Horak, & Zajac, 1999), a knee strategy is important for the termination of forward movement (Iqbal & Pai, 2000; Oddsson & Thorstensson, 1986). The results from a study involving fast forward flexion of the trunk from a quiet standing position demonstrated that the knee joint was the first to flex in order to offset the anteriorly displacing mass of the HAT segment so that a loss of balance did not occur (Oddsson et al., 1986). The authors explained the functional importance of this knee strategy as the length of the femurs allows for posterior CoM adjustment with knee flexion.

In support of these postural control strategies at the knee, Iqbal et al. utilized computer-modeling techniques to test the feasibility of a knee strategy in the control of the CoM position during the termination of forward movement (Iqbal et al., 2000). The major finding did show that knee flexion could contribute to postural control in the sagittal plane during the termination of forward movement. Thus far we have examined both abdominal (see "Core Stabilization") and lower extremity postural controls in perturbations as well as dynamic movement. The temporal sequence of activations arising from the abdomen lends insight into the organization of postural control.

The Core as the "Keystone" to Postural Control

Within the kinetic chain, studies have demonstrated a proximal to distal temporal pattern of muscle activation and biomechanical movement (Bouisset & Zattara, 1981; Zattara & Bouisset, 1988). These two studies monitored the postural responses to upper extremity motion using surface EMG to monitor muscle activation sequences and accelerometers to monitor the biomechanical responses of the shank, thigh, pelvis, and wrist. Temporal patterns of EMG activation during voluntary upper arm movement demonstrated that leg, hip, and trunk activation (tensor fascia latae, rectus femoris, and erector spinae) preceded that of deltoid onset. These EMG activation patterns were supported with the same proximal to distal sequence of local segmental accelerations. Although these studies demonstrated a proximal to distal approach to maintain postural control during upper arm movements, they did not monitor other postural muscles such as the abdominals.

The work of Hodges and colleagues demonstrates that the abdominals, both global and local are not only contributors to postural control but are at the epicenter of the temporal sequencing prior to extremity movement (Hodges et al., 1997b; Hodges et al., 1997a). As described earlier, the global abdominals (RA and EO) have been demonstrated to provide a flexor moment to the trunk when it is necessary to offset a perturbation in the body's center of gravity during sagittal plane surface translation perturbations as well as direction specific upper and lower extremity motion (Hodges et al., 2004; Hodges et al., 1997b; Hodges et al., 1997a). However, the local stabilizing component of postural control is demonstrated by earlier activation onset of the TrA relative to global abdominal activation and/or upper or lower extremity movement during reaction based tasks, as well as self-initiated and unexpected trunk loading conditions (Cresswell et al., 1994b; Hodges et al., 1997a; Hodges et al., 1997b). As TrA activation results in increases in intra-abdominal pressure (IAP), this mechanism has been demonstrated to be of prime importance in the maintenance of spinal control, lumbopelvic control, as well as overall postural control (Cresswell et al., 1994c; Hodges et al., 2004).

Core Stabilization and Kinetic Chain Function

Clinical observations of the influence of the HAT segment (primarily the lumbar spine and pelvis) on the lower extremity suggest: 1) excessive anterior pelvic tilting presents with a lordotic curvature, femoral internal rotation, knee hyperextension, and foot pronation and 2) excessive posterior pelvic tilting combines a flattened lumbar spine with external femoral rotation, knee hyperextension, and foot supination (Kendall, 1993). These clinical observations have been associated with “faulty” alignments and support the need to evaluate the influence of the more proximal segments (pelvis) on the distal lower extremity segments. To fully observe this core to lower extremity influence, we will first examine the consequences of dysfunction of the local abdominal system.

Abdominal and Kinetic Chain Dysfunction

While the etiology of TrA dysfunction is uncertain, it has been associated with low back pain populations (Ferreira et al., 2004; Hodges, 2001; Hodges et al., 1996). These studies demonstrated that activation of the TrA prior to upper limb movement was delayed in subjects with low back pain as compared to healthy controls. Subsequently, the mechanism to increase IAP via TrA activation and thus provide local spinal stiffness and lumbopelvic control was suggested to be lost. Although the authors of these studies could not substantiate whether the low back injury caused TrA dysfunction or vice versa, this connection between the two is apparent and has been reproduced (Hodges, 2001).

When the TrA is dysfunctional, it has been suggested that the lower fibers of the TrA cannot exert a horizontal force on the anterior pelvis to limit pubic symphysis shear forces, thus leading to hip adductor strains (Cowan et al., 2004). Results of this study examining the relationship between the TrA activation onset and groin pain (comparing 12 controls to 10 groin-injured subjects), demonstrated that the feedforward activation timing of the TrA prior to lower extremity movement was present in the normal healthy controls but not in the injured group. Although the relationship between the TrA dysfunction and hip adductor injury was demonstrated, which factor was the cause is still unclear. This preliminary evidence shows that as one "link in the kinetic chain" is dysfunctional, there is potential for injuries to the lower extremity.

The development of IAP, through TrA activation, is thought to provide an optimal amount of spinal stiffness, and contribute to lumbopelvic and postural control (Cresswell et al., 1994c; Hodges et al., 2004; Richardson et al., 2004). Research establishing IAP increases in preparation for landing from a jump demonstrates that local abdominal control is instrumental prior to landing in order to maintain postural control after impact (Cresswell, Blake, & Thorstensson, 1994a). This study examined if a rotational strength training program aimed at the transversus abdominis and oblique musculature could increase IAP in a variety of tasks including drop-jumping. Although IAP was developed prior to landing at base line testing, the rate of IAP development was enhanced after the 10 week program suggesting better spinal, lumbopelvic, and postural control in landing through TrA activation (Cresswell et al., 1994a). Alternatively if the TrA is dysfunctional and the biomechanical mechanisms of core stabilization are compromised, the loss of spinal, lumbopelvic, or overall postural control could lead to a loss of lumbopelvic and/or trunk control and contribute to the injurious "position of no return"(Ireland et al., 1997).

Rationale for Local Abdominal Dysfunction in Females

Sex differences in epidemiological reports of low back pain suggest that females have higher rates of low back pain than males (Nadler et al., 2002a). A potential contribution to low back pain among females may be the reported increased co-contractions of the global abdominals during isometric trunk loading conditions when compared to males (Granata et al., 2001). As the authors of this study acknowledged that higher co-contractions of the global abdominals do not ensure adequate lumbar stabilization, others have shown that excessive global muscle activation and its associated excessive trunk stiffness actually destabilizes the spinal segments (Gardner-Morse & Stokes, 2001; McGill, 2002). These findings together suggest that females may have TrA dysfunction that presents itself with global abdominal compensation.

Females with low back pain often have associated hip muscle imbalances (particularly bilateral differences in mean hip abductor/extension strength) and higher incidences of lower extremity injuries (Nadler et al., 2001; Nadler et al., 2002b; Nadler, Malanga, Deprince, Stitik, & Feinberg, 2000). These prospective studies assessing hip strength ratios suggest a decreased ability of females to utilize the hip extensors and rely more on knee and ankle musculature to absorb the GRFs during landing. Hip absorption has been reported to contribute to 7-25% of the total lower extremity joint absorption in females (Kulas et al. preparation (a))(Decker et al., 2003; Devita et al., 1992). In contrast, males absorbed 30-40% of the total lower extremity absorption at the hip (Decker et al., 2003; Zhang et al., 2000). Although these low hip absorption reports in females cannot be solely attributed to the hip muscle imbalances and low back pain reports in females, there is potential that proper core stabilization could effectively facilitate hip absorption, through increased lumbopelvic control, while decreasing the absorption requirements at the ankle and/or knee during landing activities.

While core stabilization regimens used within the sports realm focus on global abdominal strengthening, there is reason to believe that females may also need to focus on facilitating local abdominal function to provide optimal core stability. Sex differences in abdominal recruitment patterns demonstrate that females are more reliant on their global abdominals (RA and EO) than males (Granata et al., 2001). If females rely more on the global abdominals than the local abdominals (TrA and IO) during a dynamic sporting situation such as landing, loss of local spinal control, lumbopelvic control, and overall postural control (as influenced by the HAT segment) may lead to increased risk for lower extremity injuries compared to males.

Summary

The goal of this review of literature was to provide a rationale for considering the influence of core stabilization on lower extremity biomechanics during dynamic movement. This review provided a background of core stabilization that highlighted the importance of both global and local abdominal function to attain an optimally stable core. With local abdominal function providing a keystone piece to overall core stabilization, the review first focused on the role of the Transversus Abdominis during Abdominal

Hollowing, a clinical exercise aimed to enhance local abdominal function. Next, common lower extremity biomechanical measures were reviewed as they have been purported to explain a person's injury potential during sporting tasks such as landing. A focus on lower extremity energetics and leg spring stiffness provided both individual joint and whole-body measures that reflect how the GRFs are distributed and the body's response to a given GRF during landing. Lastly, while adopting a kinetic chain approach to understanding human movement, a rationale for considering the influence of core stabilization on lower extremity biomechanics was proposed. Although research seems to support an influence of core stabilization on lower extremity function, there is little information substantiating this. However, based on the available research, it seems compelling that there is a need for research to examine the influence of core stabilization on lower extremity function.

CHAPTER III

METHODS

Design

The study design followed a two-day (control and intervention days) within subject model in which two conditions on each day were compared, (control-control and control-experimental) with between sex comparisons. Subjects served as their own controls and performed two sets of double leg drop landings from a height of 60cm on two separate days. On day one (control day), both sets of double leg landings were performed to "land as naturally as you can." On day two (intervention day), subjects first performed one set of natural drop landings followed by instruction of abdominal hollowing (AH, intervention). Subjects were then retested in the drop landing task while abdominal hollowing and served as the experimental condition. The independent variables were the AH condition (within subjects) during the double leg landing task and sex (between subjects). Dependent variables collected during the landing task were: 1) Transversus abdominis-internal oblique (TrA-IO) EMG in proportion to the summed total abdominal EMG signal (RA + EO + TrA-IO) during the last 150ms prior to landing and 150ms immediately following landing, 2) leg spring stiffness (LSS), and 3) joint energetics each at the ankle, knee, and hip. The latter two variables assessed lower extremity biomechanics between control and experimental conditions and were calculated from kinetic and kinematic data acquired through a force plate interfaced with a 3-dimensional electromagnetic tracking device.

Subjects

Twenty-five males (Age=22.2±6.1yrs, Mass=81.7±11.3kg, BMI=25.5±3.3kg/m^2) and twenty-five females (Age=20.7±4.5yrs, Mass=65.4±9.4kg, BMI=23.8±3.4kg/m^2) participated in this study. All subjects qualified as recreationally active by engaging in physical activity for at least 30 minutes, three times per week. In addition to being recreationally active, males and females were activity matched based on prior experience in jumping and landing activities. Table 1 displays this data and Appendix B presents individual subject activity levels. All subjects were apparently healthy individuals having no current injuries to the lower extremity and/or low back and no past history of surgeries to the low back and/or lower extremity. All subjects gave informed consent by signing a form approved by the University's IRB.

	Recreation: Days/Week		Recreation: Hours/Day		Activity Matching: Subject Frequencies				
	Mean	SD	Mean	SD	BB	VB	Gym	Plyos	MA
Males	4.3	1.3	1.8	1.1	17	3	1	3	1
Females	3.5	1.4	1.3	0.6	17	3	1	3	1

Table 1 Recreational Levels and Activity Matching
SD=standard deviation, BB=-Basketball, VB=Volleyball, Gym=Gymnastics, Plyos=plyometrics, MA=Martial Arts (Kickboxing, Tae Kwon Do).

Instrumentation

Kinematic data for the head, thorax, pelvis, thighs, shanks, and feet were collected at 140 Hz using the Motion Monitor electromagnetic tracking system (Ascension Star

Hardware, Ascension Technology, Burlington, VT) (Motion Monitor Software, Innovative Sports Training, Chicago, IL). Two Bertec Force Plates, Type 4060-nonconducting (Bertec Corporation, Columbus, OH) acquired ground reaction forces. Motion Monitor software sampled the ground reaction forces at 1000Hz. Surface electromyographic (EMG) data for the abdominal muscles were acquired using a Myopac 2000 system (Run Technologies, Mission Viejo, CA) and the subsequent signal processed using Datapac software (Run Technologies, Mission Viejo, CA). All surface EMG was sampled at 1000Hz.

Procedures

Subjects reported to the Applied Neuromechanics Research Laboratory for the first day of data collection. Upon completion of the informed consent, and the activity information form, height and mass measurements were assessed and manually recorded. The principal investigator then demonstrated the double leg landing from a 60cm box and the subjects practiced until comfortable with the task. Instructions to every subject included: hold the hands at sides of hips with the thumbs on top of the hips and fingers pointing downward at all times, start with both feet at the edge of the box, reach straight out with the preferred leg and shift the weight of the hips forward off the box, and land on both feet at the same time. The subject's preferred leg was determined by observing which foot was most frequently used in practicing the task. Subjects were specifically instructed not to jump up or out off the box or lower the body down. After the subject was comfortable performing the task, setup for data collection followed.

Surface EMG preparation consisted of alcohol pad scrubbing of the skin to enhance surface contact with the electrode followed by placement of the electrodes at three abdominal sites. The electrode placement for the transversus abdominis/internal obliques (TrA-IO) was 2cm medial and inferior to the anterior superior iliac spine. This location has been used to represent activation profiles of the TrA and IO and has been viewed as the best surface location to represent TrA function (Marshall et al., 2003; McGill et al., 1996; Strohl et al., 1981). Surface EMG placement for rectus abdominis was 2cm lateral to the umbilicus, while external oblique electrode placement was 12cm lateral to the umbilicus (McGill et al., 1996). Submaximal voluntary isometric contractions of all three abdominal muscles were used to normalize the EMG data for between subject comparisons. Subjects were positioned supine with hips flexed to 45° and feet flat on the floor. The subjects were required to lift the feet off the floor approximately 2.5 cm and hold for 3 seconds. Three trials were collected. This procedure has been demonstrated to provide reliable submaximal activation of all the abdominal muscles simultaneously (Dankaerts et al., 2004; O'Sullivan et al., 1998). Further, it has been suggested that amplitude normalization procedures be chosen based on the nature of the abdominal activation patterns and population used (Allison, Godfrey, & Robinson, 1998). Because abdominal hollowing does not involve maximal muscle activations, normalization to the SMVICs was utilized.

Kinematic setup included the examiner attaching six-degree-of-freedom-position sensors (Ascension Technologies, Burlington, VT) with double-sided tape to the following sites: over the anterior mid-shaft of the third metatarsals, the mid-shaft of the medial tibias, and the lateral aspect of mid-shaft of the femurs. Additional sensors were attached to the sacrum, thorax at the level of C7, and the occiput of the head. Estimation of the joint centers were calculated based on the midpoint between two points on the medial and lateral aspects of the ankle and knee joints while a series of thigh positions relative to the sacrum were used to estimate the hip joint center (Leardini et al., 1999; Madigan et al., 2003).

Subjects performed five 60cm double leg landing trials while acquiring surface EMG, kinematic, and kinetic data. After this set of landings was completed, subjects rested for ten minutes. Subjects then performed ten double leg drop landings while acquiring all surface EMG, kinematic, and kinetic data. This second set of drop landings concluded the data collection for day one (control day). Subjects were then scheduled to report for a second day of data collection at least 1 but no more than 7 days later.

On day two, all subjects underwent the EMG and kinematic setup identical to day one. All subjects again performed one set of five double leg landings using the exact methods described on day one. Following the acquisition of the first five landing trials (control condition, day 2), subjects were instructed in the abdominal hollowing muscle activation pattern. Instruction was partitioned into two stages: four-point kneeling followed by a standing assessment (Hodges, Richardson, & Jull, 1996). First, in the four point kneeling position, subjects were taught to draw the lower abdomen up and in towards the spine while the examiner visually monitored for unwarranted substitutions such as lumbar flexion and shallow or absent breathing (Richardson et al., 1999). This method of instruction in the four point kneeling position has previously demonstrated

TrA as the primary abdominal muscle activated with the use of real-time ultrasound (Critchley, 2002). Visual inspection of all abdominal activity was monitored via a real-time EMG display for the examiner to ensure that specific activation of the transverse abdominis/internal oblique electrode placement occurred with minimal rectus abdominis and external oblique activation.

Following performance of AH in the four point kneeling position, subjects were positioned in standing for AH assessment. Subjects were instructed to perform AH with the same effort and intensity as in the four point kneeling position. Visual inspection of a decrease in lower abdominal circumference as well as no changes in lumbar spine flexion was monitored for by the examiner. The experimental setup to acquire abdominal electromyographic activity in the clinical four point kneeling teaching position, as well as the standing position are presented in Figure 4.

In order to ensure that subjects could perform AH in a standing position, a pilot study validated the relative abdominal contribution of the TrA-IO to the previously validated and clinically used four point kneeling position (Kulas et al. in preparation (b)). The high intraclass correlation coefficients (ICCs) and standard error of the measurements (SEM) presented in Table 2 reveal a high agreement between four point kneeling and standing position when verbal cues were present to match the intensity or effort across conditions and with a 95% confidence interval of ±6% which is a measure of the precision between conditions.

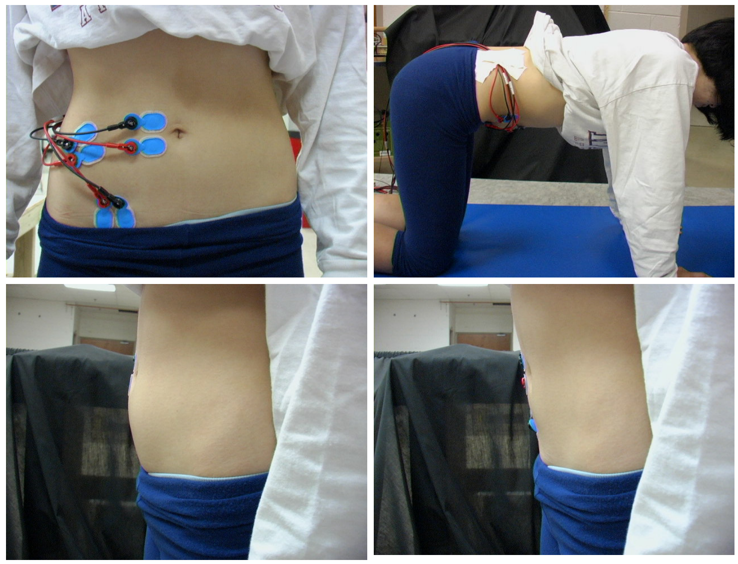

Figure 4 Abdominal EMG Setup and Clinical Positions to Instruct AH
*Clockwise from top left: Abdominal EMG placement, four-point kneeling clinical instruction position, standing with abdomen abdominal hollowed, standing relaxed abdomen.

							Partitioned Variances		
Positions	ICC 2,k	SEM	95% CI	95% Total Range	Mean 1	Mean 2	BMS eta^2	TMS eta^2	EMS eta^2
4 point kneeling & Stand Target	.93	2.8%	± 5.6%	10.4%	71%	75%	92.0%	4.0%	5.0%
4 point kneeling & Stand	.49	8.6%	± 17.4%	34.8%	79%	78%	66.0%	.2%	34.0%

Table 2 Reliability and Precision of Surface EMG during Abdominal Hollowing
Ratio measure (%) = TrA-IO (%SMVIC) / Total Abdominal Activation (RA%+EO%+TrA-IO%); N=6

Once subjects completed the AH instruction in the standing position, they performed the double leg landing task while simultaneously abdominal hollowing. Using the same kinematic setup from the first control condition, subjects performed ten double leg landing trials (experimental condition). Prior to each of these ten landing trials, while standing on top of the 60cm box and performing AH, an abdominal circumference decrease while maintaining a neutral lumbar spine was visually checked by the examiner to ensure that the subject was performing the hollowing properly and without substitution of rectus abdominis and external oblique activity. All subjects were instructed to perform AH while on top of the 60cm box with the same intensity and effort to match the previously acquired four point kneeling and standing positions. Subjects were instructed to maintain the AH as best as they could beginning on top of the box and throughout the entire landing trial until the examiner told the subject they could step off the force plates.

Both position of sensor setup and surface EMG electrode placement remained the same throughout the duration of the entire testing process to minimize extraneous error due to differences in digitization and electrode placement. However, if a surface electrode detached from the skin, the supine submaximal normalization procedure was again performed.

Data Processing and Independent & Dependent Variables

All surface electromyographic activity was band-pass filtered between 30 and 250Hz using a 4^{th} order, zero-lag Digital Butterworth filter. The filtered root mean square (RMS) with a time constant of 25ms was then integrated for the time intervals of interest (Beith et al., 2001; Allison et al., 1998). The middle 150ms from each 3-second trial was used to compute a mean 150ms integrated EMG from standing clinical instruction. This mean was used for comparison to the 150ms preactivation and 150ms post-impact integrated EMGs. The law of constant acceleration ensures that the estimated time of free fall from the box (.6m high) is 350ms, and this ensured that the subjects were not in contact with the box 150ms before initial contact. The integrated EMG signals for the rectus abdominis, external oblique, and transversus abdominis-internal obliques were imported into a spreadsheet program and normalized to a percentage of the submaximal voluntary isometric contractions (%SMVIC) (O'Sullivan et al., 1998). To ensure that the primary proportion of the AH muscle activation pattern was being produced from the TrA-IO muscles, a ratio of the TrA-IO%SMVIC to the summed total of RA%SMVIC, EO%SMVIC, and TrA-IO%SMVIC was used as the AH measure (Kulas et al., in preparation (b)). Ratios for the RA and EO were also calculated.

Kinematic and kinetic data were low passed filtered at 12 and 60Hz respectively using a 4th order, zero-lag Digital Butterworth filter. All kinematic and kinetic data were exported into a spreadsheet for calculation of the joint energetics and kinetics. All energetic and kinetic data considered for analysis were calculated from initial contact of the force plate until the body's center of mass was at its lowest position relative to the force plate. The body's center of mass position was estimated based on the summed mass of the nine segments digitized and accounted for 89% of the total body's mass (LeVeau, 1992). The landing phase was operationally defined as from initial contact with the force plate until the body's COM position reached its lowest vertical point relative to the force plate. This phase was subsequently divided into impact (first 100ms) and stabilization (100ms to end of landing phase) phases (Kulas et al., in preparation (a)).

Average Powers – Ankle, Knee, and Hip (dependent variable)

Average powers for each of the lower extremity joints were calculated by taking the integrals of the joints respective power curves during the impact and stabilization phases and then dividing the integrated values by their respective impact and stabilization phase times. The integral of the power curve when it is negative represented the amount of work done on the joints (energy absorption). Joint powers were calculated as the product of the internal joint moment times the angular velocity. Joint powers and work calculations described here have been commonly used in biomechanical research involving highly dynamic activities such as landing and running (Decker et al., 2003; Devita et al., 1992; Ferber, Davis, & Williams, 2003; Zhang et al., 2000). All average joint powers were normalized to each subject's body weight (N).

Leg Spring Stiffness (dependent variable)

Leg Spring Stiffness was calculated by dividing the peak ground reaction force by the body's center of mass displacement from initial contact to the body's minimal position relative to the force plate (Farley et al., 1999). Leg spring stiffness values were normalized to each subject's body weight (N).

Transversus Abdominis-Internal Oblique / (RA%SMVIC + EO%SMVIC% + TrA-IO%SMVIC) (dependent variable expressed as a ratio of TrA-IO activation)

The proportion of the TrA-IO muscle activity relative to the total electrical activation contributing to AH assessed the quality of the AH maneuver during four-point kneeling, standing, 150ms prior to landing, and 150ms post-landing for each double leg landing trial. Mean integrated EMG amplitudes for each trial at 150ms prior to initial contact with the force plate, and 150ms after contact were calculated from the experimental condition. The mean amplitudes for each of the three electrode placements were converted to a percentage of submaximal isometric contraction (%SMVIC) (O'Sullivan et al., 1998). These %SMVIC mean amplitudes were used to compute the ratio indicating the contribution of the TrA-IO to the AH pattern. Similarities between this ratio when AH in the standing position to the 150ms window prior to landing (during experimental condition) assessed the ability of the subjects to adequately perform the AH during both static (standing) and dynamic (150ms prior to landing) conditions. Comparison of the TrA-IO preactivation ratio across both conditions on each day (control-control and control-experimental) established whether or not subjects could increase the contribution of this muscle 150ms prior to landing following clinical instruction. Comparisons of abdominal ratios from 150ms prior to landing to 150ms post-landing were performed during the experimental condition only to assess the ability of subjects to maintain the AH activation pattern after impact.

Sex (independent variable)

Lower extremity energetics and leg spring stiffness in both control and experimental conditions were assessed between males and females. All three abdominal ratios during preactivation and impact were assessed during the experimental condition between sexes.

Statistical Analyses

Means of five landing trials from each condition were entered for statistical analysis. All five trials were used for the first condition on both days. The first five trials were used for the second control condition on day 1 to compute a mean, while the best five trials from the 10 trials in the experimental condition were used to compute a mean and entered for statistical analysis. The best five trials were chosen as the five highest TrA-IO ratios attained 150ms prior to impact. The alpha level was set a priori at $p < 0.05$ and in order to explain any observed interactions, Tukey's HSD Post Hoc Comparisons were calculated.

1. To test hypothesis 1, Intraclass Correlation Coefficient (2,k) assessed the similarity of abdominal ratios (TrA-IO / Total normalized abdominal activity) between standing and 150ms prior to landing.
2. To test hypothesis 2, two repeated measures ANOVAs with one within variable at two levels (condition: control-control (day 1) and control-experimental (day 2)) and one between variable at two levels (sex: male, females) examined differences in TrA-IO preactivation ratios as a result of performance of abdominal hollowing and between sex.
3. To test hypothesis 3, a repeated measures ANOVA with two within variables: Muscle (3 levels - rectus abdominis, external oblique, and transversus abdominis-internal oblique ratios) and Phase (2 levels – preactivation, impact) and one between variable at two levels (sex: male, female) assessed differences in activation

ratios 150ms prior to landing and 150ms post-landing during the experimental condition and between sex.

4. To test hypothesis 4, two repeated measures ANOVAs (one for each day) with one within variable (condition at two levels: control-control (day 1) and control-experimental (day 2)) and one between variable (sex: males and females) examined leg spring stiffness changes as a result of performance of abdominal hollowing and between sex.

5. To test hypothesis 5, two repeated measures ANOVAs (one for each day) with three within variables (condition at two levels: control-control (day 1) and control-experimental (day2)); average powers at three levels: ankle, knee, and hip; and phase of landing at two levels: impact and stabilization) and one between variable (sex: males and females) assessed changes in lower extremity average powers by joint (ankle, knee, and hip) and phase (impact, stabilization) from control to experimental conditions and between sex.

CHAPTER IV

RESULTS

Abdominal Hollowing Performance

Intraclass Correlation Coefficient (ICC) 2,k and Standard Error of the Measurement (SEM) were low (ICC=.58, SEM=12%; see Appendix B for SPSS outputs) for assessing the similarity of Transversus Abdominis-Internal Oblique (TrA-IO) muscle activation ratio between abdominal hollowing in standing versus abdominal hollowing during the 150ms prior to landing (experimental condition). This ratio was calculated as the proportion of TrA-IO% SMVIC (Sub-Maximum Voluntary Isometric Contraction) to the summed normalized abdominal muscle activities (RA%, EO%, TrA-IO%). While the repeated measures ANOVA assessing TrA-IO preactivation within the two control conditions and between sex (day 1) demonstrated no significant differences by condition, males exhibited significantly higher ratios than females ($F_{(1,48)}=4.56$, $P<.05$). A second repeated measures ANOVA on day 2 (control-experimental conditions) identical to the day 1 analysis demonstrated that the experimental condition yielded significantly higher TrA-IO preactivation ratios than the control ($F_{(1,48)}=71.04$, $P<.05$) with significantly higher preactivation ratios for the males compared to females ($F_{(1,48)}=5.03$, $P<.05$). The means and standard deviations for these preactivation ratios are presented in Table 2, see Appendix C for SPSS outputs).

	Day 1		Day 2	
Sex††	Control	Control	Control	Experimental
Males	62.1±12.1	61.0±12.9	60.0±14.3	71.0±11.0†
Females	53.1±15.0	53.8±14.9	54.1±12.5	61.6±12.5†

Table 3 Means +/- Standard Deviations for TrA-IO Preactivation Ratios(%)
Ratio measure (%) = TrA-IO (%SMVIC) / Total Abdominal Activation (RA%+EO%+TrA-IO%), †= experimental significantly higher than day 2 control condition, P<.001), ††=Males significantly higher than females, P<.05)

To assess the ability of both males and females to maintain the increased TrA-IO activation ratio from 150ms prior to landing to 150ms post-impact, the sex by muscle by phase repeated measures ANOVA demonstrated a three-way interaction ($F_{(2,96)}$=6.514, P<.05, see Appendix D for SPSS outputs). Tukey's HSD Post-Hoc comparisons used to identify the interaction showed that males had no significant changes in the TrA-IO and Rectus Abdominis (RA) activation ratios across phase (150ms prior to landing and 150ms post-landing), but the External Oblique (EO) ratio significantly increased post-landing (+6.9%) whereas females demonstrated a significant decrease in TrA-IO ratios (-14.6%) with a concomitant increase in EO ratios (+10.6%) across phase. In addition, while females demonstrated significantly higher EO (+5.2%) and lower TrA-IO (-9.4%) preactivation ratios than males, these sex differences continued post-impact (EO=+8.5%, TrA-IO=-20%) with the addition of higher RA ratios (+11.4%) in females. Graphs exhibiting these differences in abdominal activation ratios are presented in Figures 5 & 6.

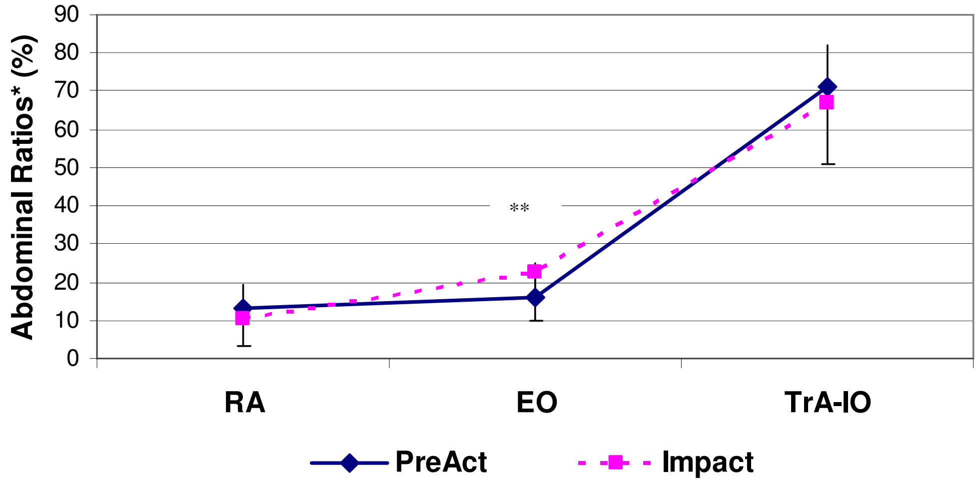

Figure 5 Changes in Abdominal EMG Ratios Across Muscle & Phase in Males**
**Muscle by Phase interaction P<.05, EO increases from preactivation to impact phase
*Ratios are computed as a relative percentage of normalized (%SMVIC) EMG: Example:RA%/(RA%+EO%+TrA-IO%); RA=Rectus Abdominis, EO=External Oblique, TrA-IO=Transversus Abdominis-Internal Oblique

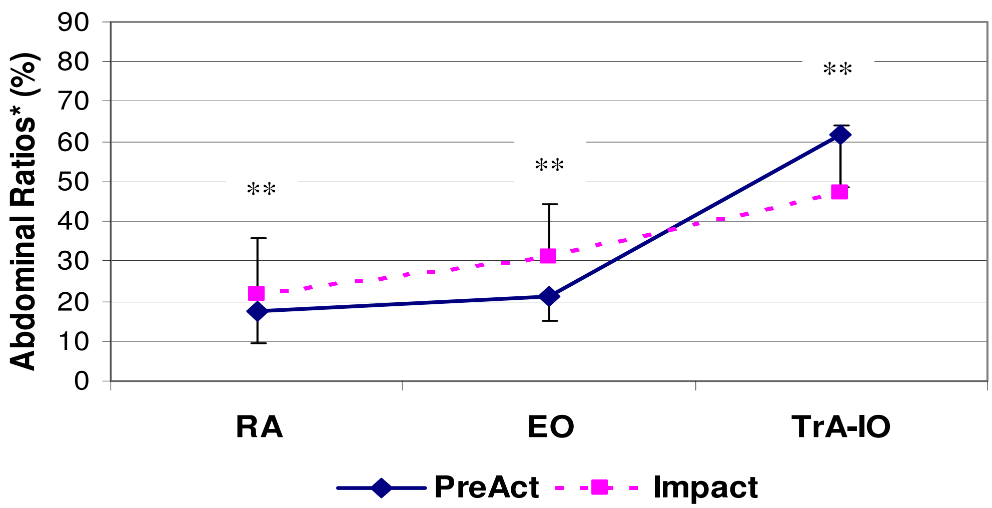

Figure 6 Changes in Abdominal EMG Ratios Across Muscle & Phase in Females**
**Muscle by Phase Interaction, P<.05; EO increases while TrA-IO decreases from preactivation to impact. *Ratios are computed as a relative percentage of normalized (%SMVIC) EMG: Example- RA%/(RA%+EO%+TrA-IO%); RA=Rectus Abdominis, EO=External Oblique, TrA-IO=Transversus Abdominis-Internal Oblique

Leg Spring Stiffness

On day 1 (control-control conditions), there was a significant main effect that demonstrated leg spring stiffness (LSS) decreased from condition 1 to 2 ($F_{(1,48)}$=42.90, P<.05). On day 2, (control-experimental conditions), there was a significant interaction observed indicating that males significantly increased LSS across conditions but females showed no change ($F_{(1,48)}$=12.064, P<.05). There were no sex differences in LSS on either day. The descriptive statistics for LSS are presented in Table 3 and the SPSS analyses can be found in Appendix E.

	Day 1†		Day 2	
	Control	Control	Control	Experimental
Males	11.4±4.9	9.7±4.4	9.0±4.0	10.0±4.4††
Females	12.5±5.2	10.8±4.4	11.6±4.4	11.1±4.3

Table 4 Means +/- Standard Deviations for Leg Spring Stiffness ((N/m)/N)
† = Decrease by condition, P<.001; †† = Sex X Condition interaction; Males significantly increase (P=.001).

Lower Extremity Energetics

On day 1, a sex by joint by phase interaction ($F_{(2,96)}$=3.28, P<.05) indicated sex differences in average powers at the knee but not the hip or ankle during the impact phase of landing. In addition, while there were no significant differences for sex or phase of landing at the hip, both the knee and ankle significantly decreased from the impact to stabilization phase in males and females. Finally, all three joints were significantly different from each other during the impact phase for males, while only the knee demonstrated significantly higher average joint powers compared to the ankle and hip during the impact phase for the female group. Figure 7 displays these relationships for day 1. Between sex differences were also observed indicating that males experienced higher rates of energy absorption overall compared to females (-.831 vs. -.581 (Nm/s)/N; $F_{(1,48)}=4.06$, $P<.05$). On day 2, a joint by phase interaction was noted ($F_{(2,96)}=42.32$, $P<.05$). Post Hoc testing revealed that while only the knee and ankle had significantly higher average powers during the impact phase of landing compared to the stabilization phase, all three joints were significantly different from one another during the impact phase with the knee producing the highest average powers and the hip showing the lowest. There were no main effects observed across condition (control-experimental) or sex. Figure 8 displays the average powers for day 2. Table 5 reports the means and standard deviations of the average joint energetics across day and the SPSS outputs can be found in Appendix F.

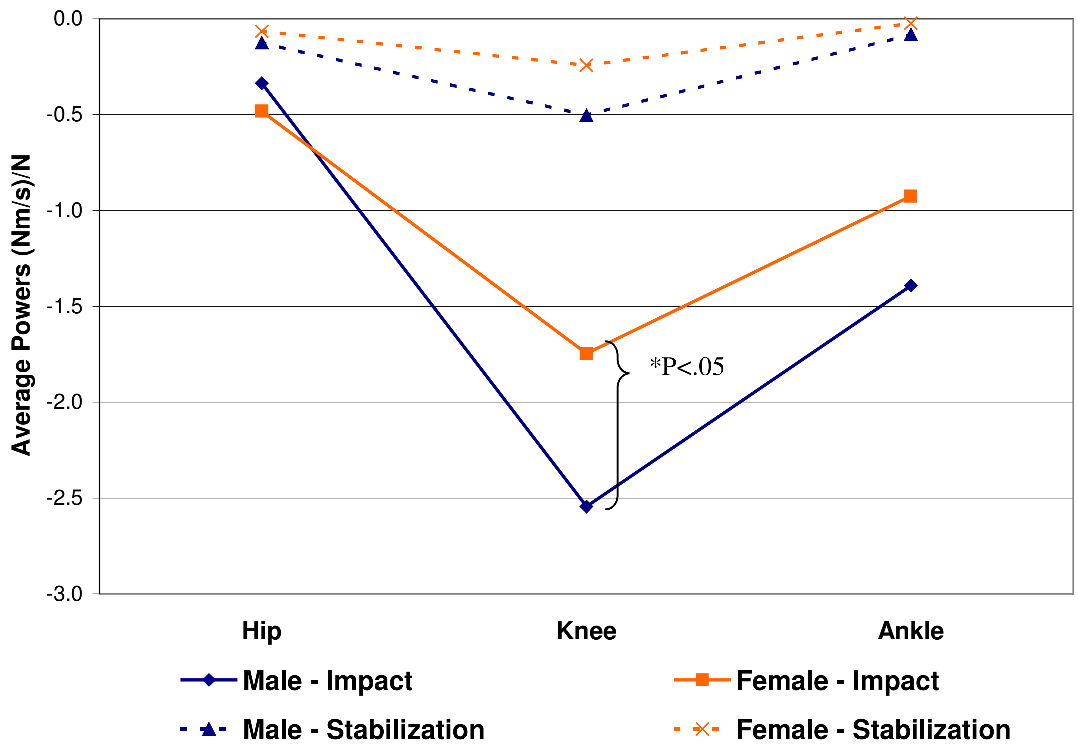

Figure 7 Average Powers across Sex, Joint, and Phase for Day 1
*= Sex*Joint*Phase Interaction; P<.05-Males significantly greater than females at knee during impact phase

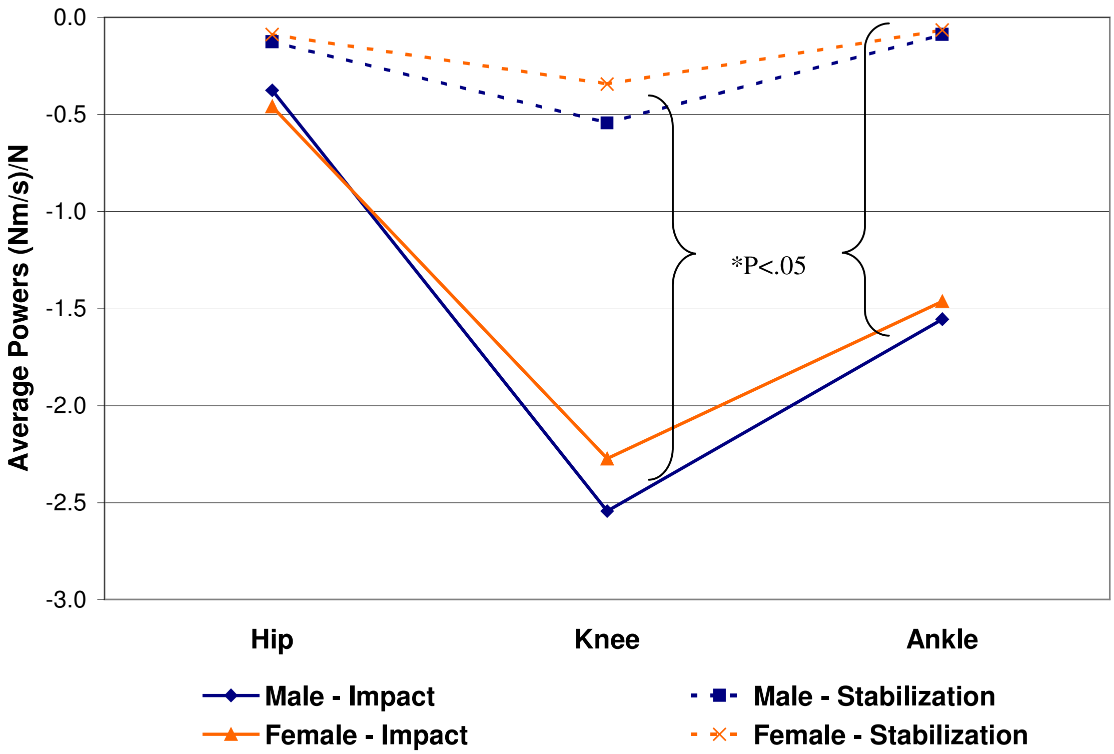

Figure 8 Average Powers across Sex, Joint, and Phase for Day 2
*Joint *Phase Interaction; P<.05, Knee and ankle at impact phase higher than stabilization phase

Sex	Joint	Landing Phase	Day 1 Mean	Day 1 SD	Day 2 Mean	Day 2 SD
Males	Hip	Impact	-.337	0.24	-.375	0.32
		Stabilization	-.126	0.10	-.125	0.12
	Knee	Impact	-2.544	1.48	-2.544	1.63
		Stabilization	-.504	0.47	-.544	0.47
	Ankle	Impact	-1.392	1.38	-1.554	1.58
		Stabilization	-.082	0.10	-.087	0.11
Females	Hip	Impact	-.482	0.31	-.459	0.36
		Stabilization	-.065	0.06	-.088	0.09
	Knee	Impact	-1.746	1.12	-2.274	1.49
		Stabilization	-.244	0.32	-.343	0.35
	Ankle	Impact	-.925	1.27	-1.463	1.32
		Stabilization	-.023	0.03	-.064	0.09

Table 5 Average Powers ((Nm/s)/N) Across Day, Joint, Landing Phase, and between Sex

CHAPTER V

DISCUSSION

The primary findings of this project indicate that although subjects could not effectively reproduce similar TrA-IO ratios from the clinical standing position to the 150ms prior to landing, all subjects did in fact demonstrate a significant increase in the TrA-IO preactivation ratio when comparing the control to experimental conditions. Males produced higher TrA-IO ratios during all control and experimental conditions. Only male subjects maintained this increased TrA-IO contraction ratio after impact. Lower extremity biomechanical analyses demonstrated that only in males did leg spring stiffness significantly increase when comparing the control to experimental condition. Although there were no significant changes in lower extremity energetics attributable to the intervention on day 2, the sex differences originally seen on day 1 - control day (males with significantly higher average powers) were not present on day 2 – intervention day.

Abdominal Activation Patterns between Sexes

The primary objective of this study was to investigate the influence of abdominal hollowing (AH) on the lower extremity biomechanics during a double leg landing task. In order to attribute any changes in lower extremity biomechanics to AH, we needed to first ensure that all subjects could effectively perform and maintain the clinically based activation pattern. Although the reliability of the TrA-IO ratio between the standing (static) and 150ms preactivation (dynamic) situation was low, all subjects demonstrated an increased ratio in the experimental condition. A plausible explanation for the lack of statistical reliability is inherent in the difference in task. During the double leg landings, the preactivation of all abdominal muscles was evident which would decrease the overall ratio of TrA-IO activity. This level of global abdominal activation was not observed during the standing AH.

Muscular preactivation has been demonstrated to be inherent during landing situations in order for the body to prepare for landing so that joint stability and postural control can be achieved (Santello & McDonagh, 1998; Santello, McDonagh, & Challis, 2001). In a series of drop landing experiments these authors demonstrated that onset-latency of lower extremity muscles were determined by drop height meaning the higher the box, the longer the latency of onset timing. In addition, the amplitude of EMG activity increased with height increases. The authors concluded that the timing and amplitude of soleus and tibialis anterior muscle preactivity is dependent on the subject's perceptions of when they will land (height of box, blind-folded etc). In applying this to our study, it may have been unreasonable to assume that the activation patterns in standing would be similar to a dynamic situation where the RA and EO muscles would normally preactivate in preparation for landing to offset any perturbations to the body's center of gravity after impact. Although not similar in amplitude to the standing position, all subjects did exhibit a significant increase in TrA-IO ratio preactivity after clinical instruction of AH. Because of RA and EO preactivity during the landing task and therefore the abdominal activation pattern was not isolated to the local abdominals, we refrain from stating that the subjects were AH during the landing but simply increased the TrA-IO ratio.

When assessing the subjects' ability to maintain this activation pattern from preactivation (150ms prior to force plate contact) to post-impact (150ms post landing), females exhibited significant increases in global abdominal ratios (RA and EO) with a significant 14% decrease in the TrA-IO ratio. Only males maintained the increased TrA-IO activation ratio throughout the landing. Because of the females' inability to maintain the augmented local abdominal activation pattern in the experimental condition post-impact, we cannot attribute any changes in the lower extremity biomechanics to the intervention. It is worth mentioning that females in this study consistently had lower TrA-IO ratios than the males in the control (53-54% vs. 60-62%) and experimental (62% vs. 71%) conditions. These results demonstrate females' reliance on the global abdominal muscles prior to and after impact. Similar to the current study Granata et al (2001) reported that females had higher RA and EO and not IO activity when compared to males in isometric trunk loading conditions (Granata et al., 2001). Based on these results the authors of this study then suggested females utilize recruitment strategies that are reliant on the global abdominals to control trunk stiffness and stability under the isometric trunk loading conditions, similar in nature to the results of the dynamic landing task used in the current study.

There are plausible explanations as to why females in this study could not maintain the increased local abdominal contraction. Subjects in this study were required to learn and immediately perform this contraction in a single testing session lasting approximately 1-1.5 hours. Although speculation, the intervention seemed to require the females to change their recruitment strategies from an even contribution by the global and local abdominals to a dominant local abdominal muscle contraction whereas males simply had to augment their preexisting local abdominal dominant recruitment patterns. Because males presented dominant local to global activation pattern in the control conditions, this intervention seemed to be easier for the males. To further demonstrate that the females could not hold a focused local activation pattern, Table 6 displays the post-impact TrA-IO ratio across all conditions and days. The female local activation ratio was at 47% during the experimental condition, demonstrating that the global abdominals comprised 53% of the post-impact activation. Although it was not a purpose to perform a statistical analysis comparing the local to global activation ratios after impact, this ratio in females demonstrates that the global abdominals contributed equal if not greater activation than the locals during all conditions. These data help to confirm that the females could not adopt the focused local abdominal intervention post-impact.

	Day 1		Day 2	
	Control	Control	Control	Experimental
Males	60.0±20.4	62.7±18.0	58.6±20.2	67.0±16.2
Females	39.6±20.2	40.8±20. 5	43.0±16.5	47.0±17.0

Table 6 Means +/- Standard Deviations for TrA-IO Post-Impact Ratios
Ratio measure (%) = TrA-IO (%SMVIC) / Total Abdominal Activation (RA%+EO%+TrA-IO%)

Given that females have been suggested to be at an increased risk for low back pain (Nadler et al., 2001) and local abdominal deficits have been observed in low back pathological populations (Ferreira et al., 2004), results from the current study suggest that females may be at increased risk for injury in the low back during dynamic tasks such as landing. Therefore the length of time needed to re-educate or alter local to global abdominal activation ratios are worth investigating for purposes of implementing local abdominal exercises in pre-season strength and conditioning programs. O'Sullivan et al (1998) conducted a ten week randomized controlled trial using 27 males and 15 females aimed at improving the contribution of the local abdominals in chronic low back pain patients (O'Sullivan et al., 1998). Results demonstrated that a ten week specific exercise protocol was sufficient to alter the local / global abdominal recruitment ratios. Although the results by sex were not reported or these measures were not assessed during dynamic landing tasks, these results have positive clinical applications because of the success in altering abdominal recruitment patterns over a ten-week period.

The consistently lower local abdominal activation ratios observed in the females may also be explained through decreased local abdominal strength. Although a limitation of our study is that muscle activation is not a measure of muscle strength, this explanation is plausible and cannot be supported nor refuted. To date we have not located a study that examined local versus global abdominal strength specifically. In the context of a landing task, it seems logical that a certain amount of local abdominal strength is necessary to maintain the intra-abdominal pressure (IAP) throughout landing. The transversus abdominis has been demonstrated to be the primary contributor to IAP development across a variety of tasks (Cresswell, 1993; Cresswell et al., 1994a; Cresswell et al., 1994c). IAP develops just prior to landing in order to control the trunk's forward motion at impact (Cresswell et al., 1994a). These authors have also demonstrated that IAP is helpful to develop a trunk extension moment during lifting and lowering (Cresswell et al., 1994c). Because of the biomechanical importance of maintaining IAP through local abdominal activation, future studies should investigate global versus local abdominal strength in at risk populations.

Sex differences in local abdominal activation may also be explained through the subject sample used. If the males participated in different activities or had higher levels of activity, this may influence our results. However by looking at Table 1 both males and females had equivalent activity participation and exercise frequency and durations. Also the body mass index (BMI), used in the abdominal EMG literature to ensure the fidelity of the abdominal signal, was equivalent across sex. Therefore, we feel that equivalent male and female samples were attained and that activity level did not contribute to a sex bias in the results of our study.

Influence of Increased Local Abdominal Activation on Lower Extremity Biomechanics

Males in this study increased and maintained the local abdominal activation ratio prior to landing and post-impact whereas the females did not. This resulted in males significantly increasing their leg spring stiffness (LSS) while females experienced no such changes during the experimental condition. Sex comparisons of LSS have previously showed that males produced higher levels of LSS during selected and preferred hopping frequencies (Granata et al., 2002). However, these sex differences were eliminated when accounting for differences in body mass as males recruited higher stiffness levels to drive their greater masses vertically. In the current study after normalizing LSS to body weight, there were no main effects for sex in LSS on either control or experimental days. Therefore the change in LSS in males that is absent in the female group can be attributed to the increased TrA-IO activation and not anthropometrics. The mechanism by which the TrA-IO influenced LSS will be described after a discussion of the contributing factors comprising LSS.

When attempting to interpret the clinical significance of LSS as it relates to this study, it must first be acknowledged that this measure is defined by the peak vertical ground reaction force (GRF) divided by the body's center of mass (CoM) displacement (Farley et al., 1998a). Its magnitude has been suggested to be linked to bony (high LSS) or soft tissue (low LSS) injuries (Butler et al., 2003). Others have attributed its meaning to the ability to develop the necessary LSS to drive the subject's body mass vertically at a selected frequency during hopping tasks (Granata et al., 2002). Control of CoM displacement is thought of as a clinically advantageous postural control mechanism in order to maintain balance during dynamic activities (Latash, 1998). So as LSS changes, the body's CoM displacement and GRF change as a result.

Sagittal joint kinematic analyses were not an original focus of this project. However when looking at individual joint angles during the landing task a condition by joint interaction ($F_{(2,48)}=6.598$, $P<.05$) demonstrated that hip total joint displacements decreased with no significant changes in knee or ankle joint displacements in the male subjects. This decrease in sagittal plane hip motion in males suggests that the hip was instrumental in controlling the HAT segment's motion. Provided that the HAT segment comprises approximately 60% of the total body's mass, the body's vertical CoM displacement decreased and thus increased LSS.

Further supporting the suggestion that hip motion most likely influenced LSS, the female subjects did not experience any changes in hip motion and LSS when comparing control to experimental conditions but did increase knee and ankle motion ($F_{(2,48)}$=10.231, $P<.05$). When describing the LSS results with total joint displacements, it provides a more informative picture of the contributing joints causing LSS to increase in the males. It seems that in males LSS was most likely increased through decreased hip motion controlling the HAT segment's CoM displacement.

Peak GRFs, the other contributor to LSS, also increased in the male subjects from the control to experimental condition (2.79BW to 2.96BW; $F_{(1,24)}$=6.25, $P<.05$). Although we did not analyze absolute joint angles at the hip, knee, and ankle, the decreased hip motion suggests a more upright posture relative to the control condition. This type of landing posture is proposed to have contributed to the increased GRF during the experimental condition. A previous study examined LSS across preferred, stiff, and soft landing techniques and found that the stiff landings had the greatest LSS values which were supported through smaller joint displacements (Kulas et al, preparation (a)). These results suggest that the subjects landed in a more erect posture at impact thus creating larger GRF and LSS values. In the current study, the GRFs increased by an average of only 6% suggesting that the GRF increase in males was probably not detrimental in terms of bony injury risk, but occurred as a result of controlling hip motion (position) in the sagittal plane relative to the GRF.

Because hip displacement decreased during the experimental condition in males, the influence of local abdominal activation to increase IAP and thus controlling lumbopelvic and postural control is plausible. As the segments proximal to the hip joint are estimated to account for over 60% of the total body's mass, the decrease in hip motion seems to have been a contributor to controlling the body's CoM displacement. Excessive forward flexion of the HAT segment has been implicated as a contributor to Anterior Cruciate Ligament injuries (Ireland et al., 1997). Therefore the ability to control the forward flexion of the HAT segment, as evidenced through the hip range of motion decreases, may then be thought of as a strategy to enhance the overall postural control and thus minimize the chances of lower extremity injuries due to the body's CoM falling outside the base of support and contributing to lower extremity injuries.

Of clinical importance to the observed LSS in males is the mechanism through which increased local abdominal activation ultimately leads to the LSS increase. Intra-abdominal pressure (IAP) increases are most closely associated with transversus abdominis (TrA) activation during lifting, lowering, and landing tasks (Cresswell et al., 1994a; Cresswell et al., 1994c). As IAP has also been demonstrated to develop in preparation for landing, it was suggested that the TrA has a preparatory function to increase IAP and thus contribute to postural control after impact (Cresswell et al., 1994a). The anatomy of the abdominal cavity extends from the diaphragm (superiorly) to the pelvic floor (inferiorly) and the abdominal wall (anteriorly) to the spine and articulating muscles (posteriorly). As IAP increases extend throughout the entire abdominal cavity, lumbar spine and pelvic motion may then be affected by changes in IAP. The high correlation between lumbar lordosis and pelvic inclination ($r=.82$, $P<.0001$) in standing radiographic measures suggest that motion in the lumbar spine influences sagittal plane pelvic motion and vice versa (Gardocki et al., 2002). Although the goal of focused TrA (Abdominal Hollowing) contractions is to stabilize the lumbar spine, pelvic motion seems to also be affected.

Although the TrA-IO ratio was maintained throughout landing and LSS increased in males there were no changes in average hip, knee, or ankle joint powers. A number of explanations need to be examined to better understand these results. As drop height was fixed to 60cm and the mass of each subject remained constant, the linear momentum (mass*velocity) prior to ground contact was theoretically fixed. At ground contact the forces applied over time in order to terminate the body's vertical momentum is referred to as an impulse. Because the impulse is a measure of the requirement to cause a change in momentum, the magnitude of the impulse is equal to the linear momentum prior to contact. This is referred to as the impulse-momentum relationship.

During the landing impulse, the hip, knee, and ankle joints act to attenuate the ground reaction forces through mechanical work. Work is done on the joints as they undergo flexion over time while producing internal extensor moments to eccentrically control the body's vertical momentum. However, mechanical work done on the lumbopelvic segment and spinal segments were not measured in this study. There is often an assumption in biomechanical modeling that the trunk segment is rigid. In reality, work is done at various spinal segments in the lumbar, thoracic and cervical spine that is most likely not adequately measured through hip absorption. This dissertation focused on increasing local abdominal activation that was proposed to influence hip, knee, and ankle energetics. However, the literature has demonstrated that TrA activation causes an increase in intervertebral stiffness at the lumbar spine (Hodges et al., 2003). This may have resulted in changes in joint energetics at the lumbar spine and not at the ankle, knee, or hip.

A limitation of the mechanical calculation of energy absorption utilized in this study is that it does not take into account energy lost due to cocontraction or energies lost as heat (Winter, 2005). In the case of landing, if a person lands with high LSS and consequently does not undergo much joint flexion due to large amounts of muscle cocontraction, this could conceivably result in relatively small joint energetics. The calculation of joint power involves the moment multiplied by the angular velocity of that joint. As angular velocities are low, the amount of physiological work done on that joint may be underestimated. Although changes in lower extremity energy absorption have been demonstrated across landing heights and landing style (stiff or soft) (Zhang et al., 2000)(Kulas et al., preparation (a)), the length of the impulses used in this study (impact and stabilization) may have been too long to detect any small changes in joint energetics due to our intervention.

A closer examination of the joint energetics across the control and experimental conditions using Cohen's D statistic for effect sizes demonstrated that the largest effect from control to experimental was 0.3, for the male hip average powers during the stabilization phase of landing. These effect sizes were calculated as the difference between two means divided by their common standard deviation. This gives us estimates of how different these means truly are. The effect size of 0.3 is low and thus only a small difference due to the intervention was detected. Again, a limiting factor to help explain these results across condition is the lack of accounting for the work done on the trunk as explained earlier.

The joint by sex by phase interaction of average power on day 1 merits further attention (Figure 7). Males exhibited higher rates of energy absorption (average powers) than females at the knee during the impact phase of landing. Decker et al (2003) demonstrated that females had higher knee and ankle energy absorption than males during this same time period (Decker et al., 2003). While there were no sex differences in hip average powers in the current study, Decker et al. (2003) reported males having higher hip energy absorption. Regardless of the sex discrepancy between the two studies, the knee was still found to be the primary energy absorber while the ankle was the secondary absorber. As others have demonstrated similar results (Zhang et al., 2000), it may be said that the knee absorption dominant findings are specific to the vertical landing task.

A potential explanation for the sex discrepancy (on day 1) among our results with those of Decker et al (2003) is that their subjects placed their arms across their chest while our study required the subjects to place the finger tips and thumbs on the greater trochanter and iliac crests. With the arms across the chest the mass of the arm segments elevate the body's center of mass position and the ensuing energetics, particularly at the hip, may have been altered as a result. Subjects in the study by Decker et al (2003) also underwent a 5-minute treadmill warm-up followed by 5-7 practice landings and then performed the landing tasks wearing standardized court shoes. Although we cannot quantify how energy absorption by the shoes themselves influenced the energetics at the ankle, knee, and hip, this methodological difference still confounds comparisons between the two studies.

The joint by sex by phase differences were absent on day 2 and there were no sex differences at the knee during the impact phase of landing (Table 7). These changes in joint energetics are not attributable to the intervention because there was no statistical difference between conditions and the effect sizes across condition in females were minimal. Because the females could not adequately maintain the increased TrA-IO ratio throughout landing, any changes in joint energetics across day are plausibly due to a learning effect.

	Unit	Day 1*	Day 2
Males	(Nm/s)/N	-.831	-.871
Females	(Nm/s)/N	-.581	-.769

Table 7 Estimated Marginal Means for Average Powers Across Day
*Males significantly higher than females, $P<.05$

Clinical Relevance of the Kinetic Chain for Injury Risk and Prevention

As augmented TrA-IO activation increased leg spring stiffness in the male subjects, a proximal muscle activation pattern influenced the distal lower extremity biomechanics. This demonstrates the presence of a kinetic chain connecting the trunk to the lower extremity and builds support for previous anecdotal observations that posture and control of the pelvis and trunk has the potential to affect the rest of the lower extremity and therefore may play a role in injury risk and prevention. This study employed a landing task where previous authors have demonstrated a distal to proximal transfer of energy absorption from the ankle to the hip (Prilutsky et al., 1994). While the intervention focused on stabilization of a more proximal segment, energy absorption occurred in a distal to proximal manner. A proximal to distal influence was demonstrated in the male subjects in this study which may have clinical implications in regards to the core as a risk factor for low back and lower extremity injury. Although we did not specifically examine the role of the local abdominal muscles as a risk factor for lower extremity injury, a discussion of the literature making these conclusions is warranted.

To gain a better understanding of the etiology of various lower extremity and low back injuries, several studies have examined anatomical abnormalities, strength deficits, or muscular recruitment differences elsewhere in the lower extremity to “link” the injury to an associated risk factor (Bullock-Saxton, 1994; Cowan et al., 2004; Ferreira et al., 2004; Hertel, Dorfman, & Braham, 2004; Loudon, Jenkins, & Loudon, 2002). For example, Loudon et al compared 20 female ACL injured subjects to 20 matched controls and examined how static postural measurements such as pelvic position, sagittal knee position, and navicular drop would compare across groups (Loudon et al., 2002). Results indicated that knee recurvatum, high navicular drop, and high subtalar pronation could adequately discriminate between the two groups. In a similar retrospective design, Hertel et al (2004) found that navicular drop and anterior pelvic tilt were significant predictors of ACL injury in both males and females (Hertel et al., 2004). In another case-control study, researchers found deficits in hip extensor recruitment in 20 subjects with a history of severe ankle sprain compared to 11 matched controls (Bullock-Saxton, 1994). And most recently, Cowan et al (2004) demonstrated the Transversus abdominis activation onset was delayed relative to the lower extremity hip adductor onset in 10 subjects with a history of groin pain compared to 12 controls (Cowan et al., 2004). While these retrospective and case control studies sought out links between deficits or pathologies and an associated injury, the biggest limitation addressed by most of these authors is that there was no way to ascertain whether the injury occurred as a result of the deficit, or the deficit contributed to the injury and was therefore a risk factor for injury.

Thus in order to most effectively evaluate injury risk, it seems logical that prospective studies are appropriate as they first evaluate measures in healthy individuals and then compare those initial measures between those who were injured and those who were not. To date, there are only a few studies that have prospectively examined the core as a risk factor for low back and lower extremity injury risk (Leetun et al., 2004; Nadler et al., 2001; Nadler et al., 2000) While Leetun et al (2004) used strength measures of trunk extension, flexion, side-bridging, hip rotation (internal and external), hip extension and hip abduction, only hip external rotation strength significantly predicted lower extremity injury (Leetun et al., 2004). Nadler and colleagues evaluated hip extension and abduction strength bilaterally on over 200 individuals and found that asymmetry in hip extensor strength was present in females reporting low back or lower extremity injuries (Nadler et al., 2000). In a follow up study to validate these observations, hip extension asymmetries were demonstrated to be a significant predictor of low back pain in the female population only (Nadler et al., 2001). Through the demonstration of proximal hip strength as a risk factor for lower extremity and low back injury, these studies demonstrate that proximal deficits at the hip may lead to lower extremity injuries.

Our study employed an intervention model to first examine a proximal factor's (increased local abdominal activation) influence on the distal hip, knee, and ankle biomechanics. Our results support a proximal to distal influence for the healthy male subjects who could adequately perform the intervention in this study whereas females could not adequately perform the intervention during the task. Females in this study demonstrated significantly lower TrA-IO ratios than males in control and experimental conditions. Rather than just assume these sex differences alone represent a risk factor for females, the females also could not maintain the increased TrA-IO contraction throughout the landing. The inability to maintain this contraction suggests that females could have a predisposition to injury due to the inability to provide adequate lumbar, lumbopelvic, and postural control through dynamic movement. As brought out through the nature of the intervention model used for this dissertation, the inability of females to maintain the augmented local abdominal recruitment strategies in a dynamic landing task suggests the need for this measure to be assessed prospectively as it may be a risk factor for low back and/or lower extremity injury.

Future Studies

The results of this dissertation support a proximal-to-distal influence in the male subjects. We demonstrated that increased local abdominal activation can change lower extremity biomechanics during a dynamic task. In order to better understand and further support this abdominal-to-lower extremity influence, several research studies are warranted.

In order to further support and explain our current findings, a proximal-to-distal assessment of muscle preactivity during a landing task is necessary. Researchers have previously demonstrated that the TrA and IO are the first muscles to activate relative to upper and/or lower extremity muscle onsets with a reaction-based model (Bouisset et al., 1981; Hodges et al., 1997a; Hodges et al., 1999b). In addition, others have reported that an increase in IAP occurs prior to landing and is suggested to serve as a postural control mechanism to control the mass of the head arms and trunk segment during landing (Cresswell et al., 1994a). This early onset of IAP coupled with studies showing that TrA is the primary contributor to IAP development (Cresswell, 1993; Cresswell et al., 1994c) suggests the functional importance of TrA preactivity in landing. To better understand how the kinetic chain prepares for landing, examination of muscle preactivation of the abdominals and lower extremity is needed.

The abdominal influence on lower extremity evaluated in this dissertation examined only sagittal plane motion in the lower extremity. As human movement and injury mechanisms involve motion in all three planes, a study assessing frontal and transverse plane lower extremity biomechanics is also needed to better understand proximal to distal biomechanical influences. Single leg landings are functional tasks commonly experienced in sports and are also commonly associated with lower extremity injuries. The intervention design employed for this dissertation using a single leg landing task would be appropriate to evaluate the kinetic chain influences in the frontal and transverse planes.

In preparation for a prospective study examining the TrA as a risk factor for low back and lower extremity injury, assessment of abdominal function during common tasks used in preparticipation physical examinations such as squatting (double and single leg) needs to be established. Real-time ultrasound imaging and surface EMG assessing the thickness of all abdominal muscles and muscle activation patterns respectively would allow for a better understanding of global versus local abdominal dominance during squatting tasks in both males and females.

A prospective study examining measures of TrA function during landing and squatting tasks would examine whether or not local abdominal function is a risk factor for low back and lower extremity injury. Measures of TrA function may include: 1) feedforward function in landing, 2) local vs. global dominance in landing and squatting, and 3) abdominal-to-gluteal relationships in landing and squatting. In addition, the validation of clinical measures such as pelvic inclination and/or lumbar lordosis that might predict local abdominal function would make it easier and more efficient for clinicians to evaluate local versus global abdominal function.

Once risk factors have been established by means of prospective studies, the next step would be to implement prevention and rehabilitation programs aimed at the restoration of adequate core function within the kinetic chain. Programs may need to start with the re-education and training of local abdominal function and then integrate this local abdominal function into a more functional task such as a squat. In order to evaluate the efficacy of these programs, prospective injury studies coupled with evaluating changes in local abdominal function in subjects not injured would be employed.

Conclusions

The results of this study support the presence of a kinetic chain where a proximal intervention (increased transversus abdominis and internal oblique ratio) influenced the distal lower extremity as assessed through leg spring stiffness. In the male participants, leg spring stiffness increased as a result of the augmented local abdominal activation pattern. Upon further investigation, hip range of motion decreases in males accounted for the decrease in the body's center of mass displacement and hence increases in leg spring stiffness. These results show that augmented local abdominal activation patterns can result in improved postural control by controlling the body's center of mass during dynamic motion.

Due to the fact that only males could adequately perform the selected intervention, our results cannot be generalized to the female participants. However, future research is needed to examine why females could not maintain the abdominal activation pattern. This inability of females to maintain the increased TrA-IO activation pattern may suggest a deficit in the kinetic chain. Finally, the proximal influence (augmented local abdominal activation patterns) on distal lower extremity biomechanics may indicate that if there is a deficit or enhancement proximally, then there may also be potential for injuries or improved joint stabilization distally.

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Appendix A: Landing Activity Questionnaire

Effect of Abdominal Hollowing on Lower Extremity Biomechanics during Double Leg Landings

Subject ID# _____________ **Date:**_____________

Age__________ **Height** __________ **Weight** _________ **Gender:** M F

Exercise Frequency: (days/week)__________ **Exercise Duration:** (hours/session)_________

1. Have you ever sustained an injury to either leg (thigh, knee, ankle, etc.)? Yes No

If so, explain____

____

2. Have you ever sustained an injury to the lower back? Yes No

If so, explain____

____

3. Have you ever been diagnosed with any low back problems? Yes No

If so, explain____

____

4. Do you have any medical conditions that prohibit you from participating in physical activity? Yes No

5. Have you in the past year participated in any team sports? If so, please list them._____

____

6. Please list types of physical and/or recreational activities typically performed.

____

Appendix B: SPSS Outputs & Syntax for Hypothesis #1

SPSS Outputs for Hypothesis #1

Within-Subjects Factors

Measure: MEASURE_1

CONDITIO	Dependent Variable
1	TRASTRA - Increased TrA in Standing
2	TRAPRA4 - TrA preactivation in exerimental condition

Descriptive Statistics

	Mean	Std. Deviation	N
TRASTRA	.75752	.185859	50
TRAPRA4	.66301	.126051	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
CONDITIO	Sphericity Assumed	.223	1	.223	16.898	.000
Error(CONDITIO)	Sphericity Assumed	.648	49	.013		

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	50.448	1	50.448	1355.507	.000
Error	1.824	49	.037		

Calculation of Intraclass Correlation Coefficient (ICC) 2,k and Standard Error of Measurement (SEM):

TMS=.223, EMS=.013, BMS=.037 SD=.186

ICC 2,K = $(BMS - EMS)/(BMS + (TMS - EMS)/50) = .583$

SEM = $SD * \sqrt{1 - ICC} = .120\%$

Syntax for Hypothesis #1

```
GLM
  trastra trapra4
  /WSFACTOR = conditio 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(conditio)
  /PRINT = DESCRIPTIVE
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = conditio .
```

Appendix C: SPSS Outputs & Syntax for Hypothesis #2

SPSS Outputs for Hypothesis #2 – Day 1

Within-Subjects Factors

Measure: MEASURE_1

COND	Dependent Variable
1	TRAPRA1= TrA/IO Preactivation - C1 - Day 1
2	TRAPRA2= TrA/IO Preactivation - C2 - Day 2

Between-Subjects Factors

		N
SEX	1 = Males	25
	2 = Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
TRAPRA1	1	.620982	.1211934	25
	2	.531003	.1500470	25
	Total	.575993	.1424315	50
TRAPRA2	1	.609578	.1289218	25
	2	.537627	.1487702	25
	Total	.573602	.1424848	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
COND	Sphericity Assumed	.000	1	.000	.070	.793	.001	.058
	Greenhouse-Geisser	.000	1.000	.000	.070	.793	.001	.058
	Huynh-Feldt	.000	1.000	.000	.070	.793	.001	.058
	Lower-bound	.000	1.000	.000	.070	.793	.001	.058
COND * SEX	Sphericity Assumed	.002	1	.002	.992	.324	.020	.164
	Greenhouse-Geisser	.002	1.000	.002	.992	.324	.020	.164
	Huynh-Feldt	.002	1.000	.002	.992	.324	.020	.164
	Lower-bound	.002	1.000	.002	.992	.324	.020	.164
Error(COND)	Sphericity Assumed	.098	48	.002				
	Greenhouse-Geisser	.098	48.000	.002				
	Huynh-Feldt	.098	48.000	.002				
	Lower-bound	.098	48.000	.002				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	33.039	1	33.039	919.553	.000	.950	1.000
SEX	.164	1	.164	4.561	.038	.087	.553
Error	1.725	48	.036				

a. Computed using alpha = .05

Estimated Marginal Means

1. SEX

Estimates

Measure: MEASURE_1

SEX	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.615	.027	.561	.669
2	.534	.027	.480	.588

Pairwise Comparisons

Measure: MEASURE_1

(I) SEX	(J) SEX	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.081*	.038	.038	.005	.157
2	1	-.081*	.038	.038	-.157	-.005

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Univariate Tests

Measure: MEASURE_1

	Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Contrast	.082	1	.082	4.561	.038	.087	4.561	.553
Error	.862	48	.018					

The F tests the effect of SEX. This test is based on the linearly independent pairwise comparisons among the estimated marginal means.

a. Computed using alpha = .05

Syntax for Hypothesis #2 – Day 1

```
GLM
  trapra1 trapra2 BY sex
  /WSFACTOR = conditio 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(sex) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(conditio) COMPARE ADJ(BONFERRONI)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = conditio
  /DESIGN = sex .
```

SPSS Outputs for Hypothesis #2 – Day 2

Within-Subjects Factors

Measure: MEASURE_1

COND	Dependent Variable
1	TRAPRA3= TrA/IO Preactivation C1 - Day 2
2	TRAPRA4= TrA/IO Preactivation C2 - Day 2

Between-Subjects Factors

		N
SEX	1 = Males	25
	2 = Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
TRAPRA3	1	.599607	.1424996	25
	2	.541303	.1254456	25
	Total	.570455	.1360910	50
TRAPRA4	1	.71012	.110328	25
	2	.61590	.125077	25
	Total	.66301	.126051	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
COND	Sphericity Assumed	.214	1	.214	71.042	.000	.597	1.000
	Greenhouse-Geisser	.214	1.000	.214	71.042	.000	.597	1.000
	Huynh-Feldt	.214	1.000	.214	71.042	.000	.597	1.000
	Lower-bound	.214	1.000	.214	71.042	.000	.597	1.000
COND * SEX	Sphericity Assumed	.008	1	.008	2.674	.109	.053	.361
	Greenhouse-Geisser	.008	1.000	.008	2.674	.109	.053	.361
	Huynh-Feldt	.008	1.000	.008	2.674	.109	.053	.361
	Lower-bound	.008	1.000	.008	2.674	.109	.053	.361
Error(COND)	Sphericity Assumed	.145	48	.003				
	Greenhouse-Geisser	.145	48.00	.003				
	Huynh-Feldt	.145	48.00	.003				
	Lower-bound	.145	48.00	.003				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	38.036	1	38.036	1315.432	.000	.965	1.000
SEX	.145	1	.145	5.028	.030	.095	.594
Error	1.388	48	.029				

a. Computed using alpha = .05

Estimated Marginal Means

1. SEX

Estimates

Measure: MEASURE_1

SEX	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.655	.024	.607	.703
2	.579	.024	.530	.627

Pairwise Comparisons

Measure: MEASURE_1

(I) SEX	(J) SEX	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	.076*	.034	.030	.008	.145
2	1	-.076*	.034	.030	-.145	-.008

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Univariate Tests

Measure: MEASURE_1

	Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Contrast	.073	1	.073	5.028	.030	.095	.594
Error	.694	48	.014				

The F tests the effect of SEX. This test is based on the linearly independent pairwise comparisons among the estimated marginal means.

a. Computed using alpha = .05

2. COND

Estimates

Measure: MEASURE_1

COND	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.570	.019	.532	.609
2	.663	.017	.629	.697

Pairwise Comparisons

Measure: MEASURE_1

(I) COND	(J) COND	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.093*	.011	.000	-.115	-.070
2	1	.093*	.011	.000	.070	.115

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Syntax for Hypothesis #2 – Day 2

```
GLM
  trapra3 trapra4 BY sex
  /WSFACTOR = cond 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(sex) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(cond) COMPARE ADJ(BONFERRONI)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = cond
  /DESIGN = sex .
```

Appendix D: SPSS Outputs & Syntax for Hypothesis #3

SPSS Outputs for Hypothesis #3

Within-Subjects Factors

Measure: MEASURE_1

MUSCLE	PHASE	Dependent Variable
1=RA	1=Preactivation	RAPRA4
	2=Impact	RAIRA4
2=EO	1=Preactivation	EOPRA4
	2=Impact	EOIRA4
3=TrA/IO	1=Preactivation	TRAPRA4
	2=Impact	TRAIRA4

Between-Subjects Factors

		N
SEX	1=Males	25
	2=Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
RAPRA4	1	.13183	.056722	25
	2	.17392	.082302	25
	Total	.15287	.073113	50
RAIRA4	1	.10339	.065528	25
	2	.21735	.140278	25
	Total	.16037	.122697	50
EOPRA4	1	.15805	.094246	25
	2	.21018	.064917	25
	Total	.18412	.084308	50
EOIRA4	1	.22683	.133642	25
	2	.31245	.133219	25
	Total	.26964	.138964	50
TRAPRA4	1	.71012	.110328	25
	2	.61590	.125077	25
	Total	.66301	.126051	50
TRAIRA4	1	.66979	.161751	25
	2	.47019	.170367	25
	Total	.56999	.192856	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
MUSCLE	Sphericity Assumed	12.274	2	6.137	185.560	.000	.794	1.000
	Greenhouse-Geisser	12.274	1.549	7.922	185.560	.000	.794	1.000
	Huynh-Feldt	12.274	1.625	7.555	185.560	.000	.794	1.000
	Lower-bound	12.274	1.000	12.274	185.560	.000	.794	1.000
MUSCLE * SEX	Sphericity Assumed	.810	2	.405	12.250	.000	.203	.995
	Greenhouse-Geisser	.810	1.549	.523	12.250	.000	.203	.983
	Huynh-Feldt	.810	1.625	.499	12.250	.000	.203	.986
	Lower-bound	.810	1.000	.810	12.250	.001	.203	.929
Error(MUSCLE)	Sphericity Assumed	3.175	96	.033				
	Greenhouse-Geisser	3.175	74.372	.043				
	Huynh-Feldt	3.175	77.988	.041				
	Lower-bound	3.175	48.000	.066				
PHASE	Sphericity Assumed	.000	1	.000	.	.	.	.
	Greenhouse-Geisser	.000	.	.	.	.	.	.
	Huynh-Feldt	.000	.	.	.	.	.	.
	Lower-bound	.000	1.000	.000	.	.	.	.
PHASE * SEX	Sphericity Assumed	.000	1	.000	.	.	.	.
	Greenhouse-Geisser	.000	.	.	.	.	.	.
	Huynh-Feldt	.000	.	.	.	.	.	.
	Lower-bound	.000	1.000	.000	.	.	.	.
Error(PHASE)	Sphericity Assumed	.000	48	.000				
	Greenhouse-Geisser	.000	.	.				
	Huynh-Feldt	.000	.	.				
	Lower-bound	.000	48.000	.000				
MUSCLE * PHASE	Sphericity Assumed	.401	2	.200	24.003	.000	.333	1.000
	Greenhouse-Geisser	.401	1.630	.246	24.003	.000	.333	1.000
	Huynh-Feldt	.401	1.715	.234	24.003	.000	.333	1.000
	Lower-bound	.401	1.000	.401	24.003	.000	.333	.998
MUSCLE * PHASE * SEX	Sphericity Assumed	.109	2	.054	6.514	.002	.119	.899
	Greenhouse-Geisser	.109	1.630	.067	6.514	.004	.119	.849
	Huynh-Feldt	.109	1.715	.063	6.514	.004	.119	.862
	Lower-bound	.109	1.000	.109	6.514	.014	.119	.706
Error(MUSCLE*PHASE)	Sphericity Assumed	.801	96	.008				
	Greenhouse-Geisser	.801	78.256	.010				
	Huynh-Feldt	.801	82.312	.010				
	Lower-bound	.801	48.000	.017				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	33.333	1	33.333	.	.	1.000	.
SEX	.000	1	.000	.	.	.	.
Error	.000	48	.000				

a. Computed using alpha = .05

Estimated Marginal Means

1. MUSCLE

Estimates

Measure: MEASURE_1

MUSCLE	Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
1	.157	.012	.133	.180
2	.227	.014	.199	.255
3	.617	.018	.580	.653

Pairwise Comparisons

Measure: MEASURE_1

(I) MUSCLE	(J) MUSCLE	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	95% Confidence Interval for Difference[a] Upper Bound
1	2	-.070*	.018	.001	-.115	-.026
	3	-.460*	.027	.000	-.528	-.392
2	1	.070*	.018	.001	.026	.115
	3	-.390*	.030	.000	-.465	-.314
3	1	.460*	.027	.000	.392	.528
	2	.390*	.030	.000	.314	.465

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

2. SEX * MUSCLE

Measure: MEASURE_1

SEX	MUSCLE	Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
1	1	.118	.016	.085	.151
	2	.192	.020	.153	.232
	3	.690	.026	.638	.742
2	1	.196	.016	.163	.229
	2	.261	.020	.222	.301
	3	.543	.026	.491	.595

3. MUSCLE * PHASE

Measure: MEASURE_1

MUSCLE	PHASE	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	1	.153	.010	.133	.173
	2	.160	.015	.129	.191
2	1	.184	.011	.161	.207
	2	.270	.019	.232	.308
3	1	.663	.017	.629	.697
	2	.570	.023	.523	.617

4. SEX * MUSCLE * PHASE

Measure: MEASURE_1

SEX	MUSCLE	PHASE	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	1	1	.132	.014	.103	.160
		2	.103	.022	.059	.147
	2	1	.158	.016	.126	.191
		2	.227	.027	.173	.280
	3	1	.710	.024	.663	.758
		2	.670	.033	.603	.737
2	1	1	.174	.014	.145	.202
		2	.217	.022	.173	.261
	2	1	.210	.016	.178	.243
		2	.312	.027	.259	.366
	3	1	.616	.024	.568	.663
		2	.470	.033	.403	.537

Syntax for Hypothesis #3

```
GLM
  rapra4 raira4 eopra4 eoira4 trapra4 traira4 BY sex
  /WSFACTOR = muscle 3 Polynomial phase 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(muscle) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(sex*muscle)
  /EMMEANS = TABLES(muscle*phase)
  /EMMEANS = TABLES(sex*muscle*phase)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = muscle phase muscle*phase
  /DESIGN = sex .
```

Appendix E: SPSS Outputs & Syntax for Hypothesis #4

SPSS Outputs for Hypothesis #4 – Day 1

Within-Subjects Factors

Measure: MEASURE_1

COND	Dependent Variable
1	LSS1= normalized leg spring stiffness - C1 - Day 1
2	LSS2= normalized leg spring stiffness - C2 - Day 1

Between-Subjects Factors

		N
SEX	1=Males	25
	2=Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
LSS1	1	11.378599	4.8843375	25
	2	12.506137	5.2266766	25
	Total	11.942368	5.0388073	50
LSS2	1	9.656297	4.4183615	25
	2	10.788554	4.4364101	25
	Total	10.222425	4.4191437	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
COND	Sphericity Assumed	73.955	1	73.955	42.900	.000	.472	1.000
	Greenhouse-Geisser	73.955	1.000	73.955	42.900	.000	.472	1.000
	Huynh-Feldt	73.955	1.000	73.955	42.900	.000	.472	1.000
	Lower-bound	73.955	1.000	73.955	42.900	.000	.472	1.000
COND * SEX	Sphericity Assumed	.000	1	.000	.000	.993	.000	.050
	Greenhouse-Geisser	.000	1.000	.000	.000	.993	.000	.050
	Huynh-Feldt	.000	1.000	.000	.000	.993	.000	.050
	Lower-bound	.000	1.000	.000	.000	.993	.000	.050
Error(COND)	Sphericity Assumed	82.747	48	1.724				
	Greenhouse-Geisser	82.747	48.000	1.724				
	Huynh-Feldt	82.747	48.000	1.724				
	Lower-bound	82.747	48.000	1.724				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	12281.952	1	12281.952	282.569	.000	.855	1.000
SEX	31.917	1	31.917	.734	.396	.015	.134
Error	2086.339	48	43.465				

a. Computed using alpha = .05

Estimated Marginal Means

COND

Estimates

Measure: MEASURE_1

COND	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	11.942	.715	10.504	13.381
2	10.222	.626	8.964	11.481

Pairwise Comparisons

Measure: MEASURE_1

(I) COND	(J) COND	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	1.720*	.263	.000	1.192	2.248
2	1	-1.720*	.263	.000	-2.248	-1.192

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Syntax for Hypothesis #4 – Day 1

```
GLM
  lss1 lss2 BY sex
  /WSFACTOR = cond 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(cond) COMPARE ADJ(BONFERRONI)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = cond
  /DESIGN = sex .
```

SPSS Outputs for Hypothesis #4: Day - 2

Within-Subjects Factors

Measure: MEASURE_1

COND	Dependent Variable
1	LSS3= normalized leg spring stiffness -C1 - Day 2
2	LSS4= normalized leg spring stiffness -C2 - Day 2

Between-Subjects Factors

		N
SEX	1=Males	25
	2=Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
LSS3	1	9.005735	3.9670932	25
	2	11.589951	4.4433986	25
	Total	10.297843	4.3683370	50
LSS4	1	9.989686	4.3761291	25
	2	11.064549	4.2674293	25
	Total	10.527117	4.3121013	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
COND	Sphericity Assumed	1.314	1	1.314	1.114	.297	.023	.179
	Greenhouse-Geisser	1.314	1.000	1.314	1.114	.297	.023	.179
	Huynh-Feldt	1.314	1.000	1.314	1.114	.297	.023	.179
	Lower-bound	1.314	1.000	1.314	1.114	.297	.023	.179
COND * SEX	Sphericity Assumed	14.238	1	14.238	12.064	.001	.201	.926
	Greenhouse-Geisser	14.238	1.000	14.238	12.064	.001	.201	.926
	Huynh-Feldt	14.238	1.000	14.238	12.064	.001	.201	.926
	Lower-bound	14.238	1.000	14.238	12.064	.001	.201	.926
Error(COND)	Sphericity Assumed	56.650	48	1.180				
	Greenhouse-Geisser	56.650	48.000	1.180				
	Huynh-Feldt	56.650	48.000	1.180				
	Lower-bound	56.650	48.000	1.180				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	10841.974	1	10841.974	307.649	.000	.865	1.000
SEX	83.680	1	83.680	2.374	.130	.047	.327
Error	1691.584	48	35.241				

a. Computed using alpha = .05

2. SEX * COND

Measure: MEASURE_1

SEX	COND	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
1	1	9.006	.842	7.312	10.699
	2	9.990	.864	8.252	11.728
2	1	11.590	.842	9.896	13.284
	2	11.065	.864	9.327	12.803

Syntax for Hypothesis #4 – Day 2

```
GLM
  lss3 lss4 BY sex
  /WSFACTOR = cond 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(sex*cond)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = cond
  /DESIGN = sex .
```

Appendix F: SPSS Outputs & Syntax for Hypothesis #5

SPSS Outputs for Hypothesis #5 – Day 1

Within-Subjects Factors

Measure: MEASURE_1

COND	JOINT	PHASE	Dependent Variable
1	1=Hip	1=Impact	HEAI1
		2=Stabilization	HEAS1
	2=Knee	1=Impact	KEAI1
		2=Stabilization	KEAS1
	3=Ankle	1=Impact	AEAI1
		2=Stabilization	AEAS1
2	1=Hip	1=Impact	HEAI2
		2=Stabilization	HEAS2
	2=Knee	1=Impact	KEAI2
		2=Stabilization	KEAS2
	3=Ankle	1=Impact	AEAI2
		2=Stabilization	AEAS2

Between-Subjects Factors

		N
SEX	1=Males	25
	2=Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
HEAI1	1	-.330605	.2428593	25
	2	-.464258	.2823594	25
	Total	-.397432	.2692494	50
HEAS1	1	-.128338	.0885631	25
	2	-.068600	.0615116	25
	Total	-.098469	.0812729	50
KEAI1	1	-2.584791	1.4482010	25
	2	-1.862533	1.2487500	25
	Total	-2.223662	1.3871172	50
KEAS1	1	-.523649	.4323119	25
	2	-.251032	.3166265	25
	Total	-.387340	.3995024	50
AEAI1	1	-1.597545	1.5883991	25
	2	-.878850	1.0535626	25
	Total	-1.238198	1.3824606	50
AEAS1	1	-.097414	.1153214	25
	2	-.024782	.0351250	25
	Total	-.061098	.0919992	50
HEAI2	1	-.343828	.2387430	25
	2	-.499011	.3420196	25
	Total	-.421419	.3022513	50
HEAS2	1	-.122851	.1011337	25
	2	-.060558	.0527326	25
	Total	-.091705	.0857994	50
KEAI2	1	-2.502543	1.5103994	25
	2	-1.630303	.9818591	25
	Total	-2.066423	1.3355311	50
KEAS2	1	-.484047	.5019665	25
	2	-.236563	.3184495	25
	Total	-.360305	.4344064	50
AEAI2	1	-1.185573	1.1640352	25
	2	-.970844	1.4929795	25
	Total	-1.078208	1.3293510	50
AEAS2	1	-.066290	.0806471	25
	2	-.020419	.0334962	25
	Total	-.043354	.0653600	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power [a]
COND	Sphericity Assumed	.495	1	.495	2.178	.147	.043	.304
	Greenhouse-Geisser	.495	1.000	.495	2.178	.147	.043	.304
	Huynh-Feldt	.495	1.000	.495	2.178	.147	.043	.304
	Lower-bound	.495	1.000	.495	2.178	.147	.043	.304
COND * SEX	Sphericity Assumed	.188	1	.188	.827	.368	.017	.145
	Greenhouse-Geisser	.188	1.000	.188	.827	.368	.017	.145
	Huynh-Feldt	.188	1.000	.188	.827	.368	.017	.145
	Lower-bound	.188	1.000	.188	.827	.368	.017	.145
Error(COND)	Sphericity Assumed	10.916	48	.227				
	Greenhouse-Geisser	10.916	48.000	.227				
	Huynh-Feldt	10.916	48.000	.227				
	Lower-bound	10.916	48.000	.227				
JOINT	Sphericity Assumed	104.466	2	52.233	55.935	.000	.538	1.000
	Greenhouse-Geisser	104.466	1.765	59.176	55.935	.000	.538	1.000
	Huynh-Feldt	104.466	1.866	55.989	55.935	.000	.538	1.000
	Lower-bound	104.466	1.000	104.466	55.935	.000	.538	1.000
JOINT * SEX	Sphericity Assumed	8.145	2	4.073	4.361	.015	.083	.743
	Greenhouse-Geisser	8.145	1.765	4.614	4.361	.019	.083	.703
	Huynh-Feldt	8.145	1.866	4.365	4.361	.018	.083	.721
	Lower-bound	8.145	1.000	8.145	4.361	.042	.083	.535
Error(JOINT)	Sphericity Assumed	89.646	96	.934				
	Greenhouse-Geisser	89.646	84.736	1.058				
	Huynh-Feldt	89.646	89.560	1.001				
	Lower-bound	89.646	48.000	1.868				
PHASE	Sphericity Assumed	169.765	1	169.765	136.559	.000	.740	1.000
	Greenhouse-Geisser	169.765	1.000	169.765	136.559	.000	.740	1.000
	Huynh-Feldt	169.765	1.000	169.765	136.559	.000	.740	1.000
	Lower-bound	169.765	1.000	169.765	136.559	.000	.740	1.000
PHASE * SEX	Sphericity Assumed	2.277	1	2.277	1.832	.182	.037	.264
	Greenhouse-Geisser	2.277	1.000	2.277	1.832	.182	.037	.264
	Huynh-Feldt	2.277	1.000	2.277	1.832	.182	.037	.264
	Lower-bound	2.277	1.000	2.277	1.832	.182	.037	.264
Error(PHASE)	Sphericity Assumed	59.672	48	1.243				
	Greenhouse-Geisser	59.672	48.000	1.243				
	Huynh-Feldt	59.672	48.000	1.243				
	Lower-bound	59.672	48.000	1.243				
COND * JOINT	Sphericity Assumed	.328	2	.164	1.066	.348	.022	.232
	Greenhouse-Geisser	.328	1.814	.181	1.066	.344	.022	.222
	Huynh-Feldt	.328	1.921	.171	1.066	.346	.022	.228
	Lower-bound	.328	1.000	.328	1.066	.307	.022	.173
COND * JOINT * SEX	Sphericity Assumed	.742	2	.371	2.413	.095	.048	.476
	Greenhouse-Geisser	.742	1.814	.409	2.413	.101	.048	.451
	Huynh-Feldt	.742	1.921	.386	2.413	.097	.048	.466
	Lower-bound	.742	1.000	.742	2.413	.127	.048	.331
Error(COND*JOINT)	Sphericity Assumed	14.758	96	.154				
	Greenhouse-Geisser	14.758	87.081	.169				
	Huynh-Feldt	14.758	92.194	.160				
	Lower-bound	14.758	48.000	.307				
COND * PHASE	Sphericity Assumed	.243	1	.243	1.590	.213	.032	.235
	Greenhouse-Geisser	.243	1.000	.243	1.590	.213	.032	.235
	Huynh-Feldt	.243	1.000	.243	1.590	.213	.032	.235
	Lower-bound	.243	1.000	.243	1.590	.213	.032	.235
COND * PHASE * SEX	Sphericity Assumed	.111	1	.111	.724	.399	.015	.133
	Greenhouse-Geisser	.111	1.000	.111	.724	.399	.015	.133
	Huynh-Feldt	.111	1.000	.111	.724	.399	.015	.133
	Lower-bound	.111	1.000	.111	.724	.399	.015	.133
Error(COND*PHASE)	Sphericity Assumed	7.346	48	.153				
	Greenhouse-Geisser	7.346	48.000	.153				
	Huynh-Feldt	7.346	48.000	.153				
	Lower-bound	7.346	48.000	.153				
JOINT * PHASE	Sphericity Assumed	53.196	2	26.598	44.368	.000	.480	1.000
	Greenhouse-Geisser	53.196	1.816	29.286	44.368	.000	.480	1.000
	Huynh-Feldt	53.196	1.923	27.660	44.368	.000	.480	1.000
	Lower-bound	53.196	1.000	53.196	44.368	.000	.480	1.000
JOINT * PHASE * SEX	Sphericity Assumed	3.933	2	1.967	3.281	.042	.064	.611
	Greenhouse-Geisser	3.933	1.816	2.165	3.281	.047	.064	.581
	Huynh-Feldt	3.933	1.923	2.045	3.281	.044	.064	.599
	Lower-bound	3.933	1.000	3.933	3.281	.076	.064	.427
Error(JOINT*PHASE)	Sphericity Assumed	57.550	96	.599				
	Greenhouse-Geisser	57.550	87.187	.660				
	Huynh-Feldt	57.550	92.313	.623				
	Lower-bound	57.550	48.000	1.199				
COND * JOINT * PHASE	Sphericity Assumed	.233	2	.117	.914	.404	.019	.204
	Greenhouse-Geisser	.233	1.578	.148	.914	.385	.019	.184
	Huynh-Feldt	.233	1.656	.141	.914	.389	.019	.188
	Lower-bound	.233	1.000	.233	.914	.344	.019	.155
COND * JOINT * PHASE * SEX	Sphericity Assumed	.698	2	.349	2.736	.070	.054	.529
	Greenhouse-Geisser	.698	1.578	.443	2.736	.083	.054	.466
	Huynh-Feldt	.698	1.656	.422	2.736	.081	.054	.478
	Lower-bound	.698	1.000	.698	2.736	.105	.054	.368
Error(COND*JOINT*PHASE)	Sphericity Assumed	12.253	96	.128				
	Greenhouse-Geisser	12.253	75.720	.162				
	Huynh-Feldt	12.253	79.488	.154				
	Lower-bound	12.253	48.000	.255				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	298.752	1	298.752	129.52	.000	.730	1.000
SEX	9.373	1	9.373	4.064	.049	.078	.506
Error	110.714	48	2.307				

a. Computed using alpha = .05

Estimated Marginal Means

1. SEX

Estimates

Measure: MEASURE_1

SEX	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-.831	.088	-1.007	-.654
2	-.581	.088	-.757	-.404

Pairwise Comparisons

Measure: MEASURE_1

(I) SEX	(J) SEX	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-.250*	.124	.049	-.499	-.001
2	1	.250*	.124	.049	.001	.499

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

Univariate Tests

Measure: MEASURE_1

	Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Contrast	.781	1	.781	4.064	.049	.078	.506
Error	9.226	48	.192				

The F tests the effect of SEX. This test is based on the linearly independent pairwise comparisons among the estimated marginal means.

a. Computed using alpha = .05

2. JOINT

Estimates

Measure: MEASURE_1

JOINT	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	-.252	.020	-.292	-.212
2	-1.259	.111	-1.483	-1.036
3	-.605	.090	-.786	-.424

Pairwise Comparisons

Measure: MEASURE_1

(I) JOINT	(J) JOINT	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	Upper Bound
1	2	1.007*	.112	.000	.728	1.286
	3	.353*	.092	.001	.125	.581
2	1	-1.007*	.112	.000	-1.286	-.728
	3	-.654*	.083	.000	-.861	-.448
3	1	-.353*	.092	.001	-.581	-.125
	2	.654*	.083	.000	.448	.861

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

6. PHASE

Estimates

Measure: MEASURE_1

PHASE	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	-1.238	.107	-1.452	-1.023
2	-.174	.021	-.217	-.131

Pairwise Comparisons

Measure: MEASURE_1

(I) PHASE	(J) PHASE	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a] Lower Bound	Upper Bound
1	2	-1.064*	.091	.000	-1.247	-.881
2	1	1.064*	.091	.000	.881	1.247

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

3. SEX * JOINT

Measure: MEASURE_1

SEX	JOINT	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	1	-.231	.028	-.288	-.175
	2	-1.524	.157	-1.840	-1.208
	3	-.737	.127	-.993	-.481
2	1	-.273	.028	-.330	-.217
	2	-.995	.157	-1.311	-.679
	3	-.474	.127	-.730	-.218

4. JOINT * PHASE

Measure: MEASURE_1

JOINT	PHASE	Mean	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
1	1	-.409	.037	-.485	-.334
	2	-.095	.011	-.116	-.074
2	1	-2.145	.177	-2.500	-1.790
	2	-.374	.054	-.483	-.265
3	1	-1.158	.174	-1.509	-.807
	2	-.052	.010	-.072	-.033

5. SEX * JOINT * PHASE

Measure: MEASURE_1

SEX	JOINT	PHASE	Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
1	1	1	-.337	.053	-.444	-.231
		2	-.126	.015	-.156	-.096
	2	1	-2.544	.250	-3.046	-2.042
		2	-.504	.077	-.658	-.350
	3	1	-1.392	.247	-1.888	-.895
		2	-.082	.014	-.109	-.054
2	1	1	-.482	.053	-.588	-.375
		2	-.065	.015	-.095	-.035
	2	1	-1.746	.250	-2.248	-1.244
		2	-.244	.077	-.398	-.090
	3	1	-.925	.247	-1.421	-.429
		2	-.023	.014	-.050	.005

SPSS Syntax for Hypothesis #5 – Day 1

```
GLM
  heai1 heas1 keai1 keas1 aeai1 aeas1 heai2 heas2 keai2 keas2 aeai2
aeas2 BY
  sex
  /WSFACTOR = cond 2 Polynomial joint 3 Polynomial phase 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(sex) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(joint) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(sex*joint)
  /EMMEANS = TABLES(joint*phase)
  /EMMEANS = TABLES(sex*joint*phase)
  /EMMEANS = TABLES(phase) COMPARE ADJ(BONFERRONI)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = cond joint phase cond*joint cond*phase joint*phase
cond*joint
 *phase
  /DESIGN = sex .
```

SPSS Outputs for Hypothesis #5 – Day 2

Within-Subjects Factors

Measure: MEASURE_1

COND	JOINT	PHASE	Dependent Variable
1	1=Hip	1=Impact	HEAI3
		2=Stabilization	HEAS3
	2=Knee	1=Impact	KEAI3
		2=Stabilization	KEAS3
	3=Ankle	1=Impact	AEAI3
		2=Stabilization	AEAS3
2	1=Hip	1=Impact	HEAI4
		2=Stabilization	HEAS4
	2=Knee	1=Impact	KEAI4
		2=Stabilization	KEAS4
	3=Ankle	1=Impact	AEAI4
		2=Stabilization	AEAS4

Between-Subjects Factors

		N
SEX	1=Males	25
	2=Females	25

Descriptive Statistics

	SEX	Mean	Std. Deviation	N
HEAI3	1	-.343544	.2701518	25
	2	-.433782	.3644947	25
	Total	-.388663	.3207742	50
HEAS3	1	-.143067	.1397789	25
	2	-.100308	.1030654	25
	Total	-.121687	.1234463	50
KEAI3	1	-2.544408	1.4924551	25
	2	-2.233636	1.4141091	25
	Total	-2.389022	1.4474346	50
KEAS3	1	-.520348	.4421865	25
	2	-.325986	.2869119	25
	Total	-.423167	.3817399	50
AEAI3	1	-1.606983	1.4914036	25
	2	-1.436182	1.3219423	25
	Total	-1.521583	1.3974339	50
AEAS3	1	-.084553	.0890316	25
	2	-.069152	.0846191	25
	Total	-.076852	.0863138	50
HEAI4	1	-.407015	.3653679	25
	2	-.484382	.3474789	25
	Total	-.445698	.3550357	50
HEAS4	1	-.106414	.1052927	25
	2	-.075490	.0829881	25
	Total	-.090952	.0951175	50
KEAI4	1	-2.543480	1.7556091	25
	2	-2.313382	1.5724179	25
	Total	-2.428431	1.6535286	50
KEAS4	1	-.567947	.5041543	25
	2	-.359885	.4074089	25
	Total	-.463916	.4656531	50
AEAI4	1	-1.488863	1.6736256	25
	2	-1.339190	1.3177217	25
	Total	-1.414027	1.4926893	50
AEAS4	1	-.090093	.1251026	25
	2	-.059514	.0892611	25
	Total	-.074803	.1086583	50

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power [a]
COND	Sphericity Assumed	4.130E-05	1	4.13E-05	.000	.989	.000	.050
	Greenhouse-Geisser	4.130E-05	1.000	4.13E-05	.000	.989	.000	.050
	Huynh-Feldt	4.130E-05	1.000	4.13E-05	.000	.989	.000	.050
	Lower-bound	4.130E-05	1.000	4.13E-05	.000	.989	.000	.050
COND * SEX	Sphericity Assumed	.005	1	.005	.027	.870	.001	.053
	Greenhouse-Geisser	.005	1.000	.005	.027	.870	.001	.053
	Huynh-Feldt	.005	1.000	.005	.027	.870	.001	.053
	Lower-bound	.005	1.000	.005	.027	.870	.001	.053
Error(COND)	Sphericity Assumed	9.547	48	.199				
	Greenhouse-Geisser	9.547	48.000	.199				
	Huynh-Feldt	9.547	48.000	.199				
	Lower-bound	9.547	48.000	.199				
JOINT	Sphericity Assumed	136.273	2	68.136	54.099	.000	.530	1.000
	Greenhouse-Geisser	136.273	1.676	81.308	54.099	.000	.530	1.000
	Huynh-Feldt	136.273	1.766	77.172	54.099	.000	.530	1.000
	Lower-bound	136.273	1.000	136.273	54.099	.000	.530	1.000
JOINT * SEX	Sphericity Assumed	1.688	2	.844	.670	.514	.014	.160
	Greenhouse-Geisser	1.688	1.676	1.007	.670	.489	.014	.150
	Huynh-Feldt	1.688	1.766	.956	.670	.496	.014	.153
	Lower-bound	1.688	1.000	1.688	.670	.417	.014	.126
Error(JOINT)	Sphericity Assumed	120.909	96	1.259				
	Greenhouse-Geisser	120.909	80.448	1.503				
	Huynh-Feldt	120.909	84.760	1.426				
	Lower-bound	120.909	48.000	2.519				
PHASE	Sphericity Assumed	224.240	1	224.240	133.189	.000	.735	1.000
	Greenhouse-Geisser	224.240	1.000	224.240	133.189	.000	.735	1.000
	Huynh-Feldt	224.240	1.000	224.240	133.189	.000	.735	1.000
	Lower-bound	224.240	1.000	224.240	133.189	.000	.735	1.000
PHASE * SEX	Sphericity Assumed	.031	1	.031	.018	.893	.000	.052
	Greenhouse-Geisser	.031	1.000	.031	.018	.893	.000	.052
	Huynh-Feldt	.031	1.000	.031	.018	.893	.000	.052
	Lower-bound	.031	1.000	.031	.018	.893	.000	.052
Error(PHASE)	Sphericity Assumed	80.814	48	1.684				
	Greenhouse-Geisser	80.814	48.000	1.684				
	Huynh-Feldt	80.814	48.000	1.684				
	Lower-bound	80.814	48.000	1.684				
COND * JOINT	Sphericity Assumed	.239	2	.120	.670	.514	.014	.160
	Greenhouse-Geisser	.239	1.745	.137	.670	.495	.014	.152
	Huynh-Feldt	.239	1.843	.130	.670	.503	.014	.155
	Lower-bound	.239	1.000	.239	.670	.417	.014	.126
COND * JOINT * SEX	Sphericity Assumed	.009	2	.004	.025	.976	.001	.054
	Greenhouse-Geisser	.009	1.745	.005	.025	.964	.001	.053
	Huynh-Feldt	.009	1.843	.005	.025	.969	.001	.053
	Lower-bound	.009	1.000	.009	.025	.876	.001	.053
Error(COND*JOINT)	Sphericity Assumed	17.129	96	.178				
	Greenhouse-Geisser	17.129	83.739	.205				
	Huynh-Feldt	17.129	88.442	.194				
	Lower-bound	17.129	48.000	.357				
COND * PHASE	Sphericity Assumed	.002	1	.002	.012	.912	.000	.051
	Greenhouse-Geisser	.002	1.000	.002	.012	.912	.000	.051
	Huynh-Feldt	.002	1.000	.002	.012	.912	.000	.051
	Lower-bound	.002	1.000	.002	.012	.912	.000	.051
COND * PHASE * SEX	Sphericity Assumed	.012	1	.012	.094	.760	.002	.060
	Greenhouse-Geisser	.012	1.000	.012	.094	.760	.002	.060
	Huynh-Feldt	.012	1.000	.012	.094	.760	.002	.060
	Lower-bound	.012	1.000	.012	.094	.760	.002	.060
Error(COND*PHASE)	Sphericity Assumed	5.953	48	.124				
	Greenhouse-Geisser	5.953	48.000	.124				
	Huynh-Feldt	5.953	48.000	.124				
	Lower-bound	5.953	48.000	.124				
JOINT * PHASE	Sphericity Assumed	70.569	2	35.285	42.320	.000	.469	1.000
	Greenhouse-Geisser	70.569	1.972	35.778	42.320	.000	.469	1.000
	Huynh-Feldt	70.569	2.000	35.285	42.320	.000	.469	1.000
	Lower-bound	70.569	1.000	70.569	42.320	.000	.469	1.000
JOINT * PHASE * SEX	Sphericity Assumed	.447	2	.223	.268	.766	.006	.091
	Greenhouse-Geisser	.447	1.972	.226	.268	.763	.006	.091
	Huynh-Feldt	.447	2.000	.223	.268	.766	.006	.091
	Lower-bound	.447	1.000	.447	.268	.607	.006	.080
Error(JOINT*PHASE)	Sphericity Assumed	80.041	96	.834				
	Greenhouse-Geisser	80.041	94.677	.845				
	Huynh-Feldt	80.041	96.000	.834				
	Lower-bound	80.041	48.000	1.668				
COND * JOINT * PHASE	Sphericity Assumed	.234	2	.117	.909	.406	.019	.203
	Greenhouse-Geisser	.234	1.628	.144	.909	.389	.019	.186
	Huynh-Feldt	.234	1.713	.137	.909	.393	.019	.190
	Lower-bound	.234	1.000	.234	.909	.345	.019	.154
COND * JOINT * PHASE * SEX	Sphericity Assumed	.022	2	.011	.086	.918	.002	.063
	Greenhouse-Geisser	.022	1.628	.014	.086	.881	.002	.062
	Huynh-Feldt	.022	1.713	.013	.086	.891	.002	.062
	Lower-bound	.022	1.000	.022	.086	.770	.002	.060
Error(COND*JOINT*PHASE)	Sphericity Assumed	12.350	96	.129				
	Greenhouse-Geisser	12.350	78.160	.158				
	Huynh-Feldt	12.350	82.205	.150				
	Lower-bound	12.350	48.000	.257				

a. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Observed Power[a]
Intercept	403.342	1	403.342	136.226	.000	.739	1.000
SEX	1.540	1	1.540	.520	.474	.011	.109
Error	142.119	48	2.961				

a. Computed using alpha = .05

Estimated Marginal Means

2. JOINT

Estimates

Measure: MEASURE_1

JOINT	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-.262	.026	-.315	-.209
2	-1.426	.128	-1.684	-1.168
3	-.772	.101	-.975	-.569

Pairwise Comparisons

Measure: MEASURE_1

(I) JOINT	(J) JOINT	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	1.164*	.135	.000	.831	1.498
	3	.510*	.102	.000	.258	.762
2	1	-1.164*	.135	.000	-1.498	-.831
	3	-.654*	.097	.000	-.894	-.414
3	1	-.510*	.102	.000	-.762	-.258
	2	.654*	.097	.000	.414	.894

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

3. PHASE

Estimates

Measure: MEASURE_1

PHASE	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-1.431	.122	-1.677	-1.185
2	-.209	.023	-.254	-.163

Pairwise Comparisons

Measure: MEASURE_1

(I) PHASE	(J) PHASE	Mean Difference (I-J)	Std. Error	Sig.[a]	95% Confidence Interval for Difference[a]	
					Lower Bound	Upper Bound
1	2	-1.223*	.106	.000	-1.436	-1.010
2	1	1.223*	.106	.000	1.010	1.436

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

a. Adjustment for multiple comparisons: Bonferroni.

4. JOINT * PHASE

Measure: MEASURE_1

JOINT	PHASE	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
1	1	-.417	.045	-.508	-.326
	2	-.106	.013	-.133	-.079
2	1	-2.409	.212	-2.834	-1.983
	2	-.444	.057	-.558	-.330
3	1	-1.468	.195	-1.859	-1.076
	2	-.076	.013	-.102	-.049

SPSS Syntax for Hypothesis #5 – Day 2

```
GLM
  heai3 heas3 keai3 keas3 aeai3 aeas3 heai4 heas4 keai4 keas4 aeai4
aeas4 BY
  sex
  /WSFACTOR = cond 2 Polynomial joint 3 Polynomial phase 2 Polynomial
  /METHOD = SSTYPE(3)
  /EMMEANS = TABLES(sex) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(joint) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(phase) COMPARE ADJ(BONFERRONI)
  /EMMEANS = TABLES(joint*phase)
  /EMMEANS = TABLES(sex*joint*phase)
  /PRINT = DESCRIPTIVE ETASQ OPOWER
  /CRITERIA = ALPHA(.05)
  /WSDESIGN = cond joint phase cond*joint cond*phase joint*phase
cond*joint
 *phase
  /DESIGN = sex .
```

Appendix G: Power Analysis to Estimate Sample Size

To estimate total sample size needed to achieve a power of .80, effect sizes of leg spring stiffness were used from previously acquired data between males and females that utilized an abdominal hollowing condition (Kulas, Windley, & Schmitz, *2005*). Of all the variables of interest to the hypotheses of this dissertation, leg spring stiffness had the lowest effect size and was therefore used for the power analysis.

$$d = \frac{\mu_1 - \mu_2}{\sigma_{X1-X2}} = \frac{77.38 - 44.76}{59.85} = .56\,;\; n = 2\left(\frac{\delta}{d}\right)^2 = 2\left(\frac{2.80}{.56}\right)^2 = 49.69 = 50$$

Appendix H: Activity Matching Spreadsheet

Subject	Sex	Age	Height (m)	Mass (kg)	BMI kg/m^2	Rec Act days/wk	Rec Act hrs/day	Subject Match #	Sport Chosen	Landing Act < 6 Months (1)	Landing Act < 6 Months (2)	Landing Act > Six Months (1)	Landing Act > Six Months (2)
1	1	21	1.783	77.7	24.44	7.0	2.0	2	1	1		1	
5	1	21	1.84	81.3	24.01	3.0	2.0	11	5	1	5	5	
9	1	22	1.77	76.8	24.51	4.0	1.5	3	1	1			
12	1	35	1.81	93.3	28.48	3.0	1.0	6	1	1		1	
14	1	20	1.82	82.9	25.03	6.0	1.5	26	5	1	5	1	
15	1	24	1.83	79.1	23.62	2.0	1.0	7	1	1		1	
16	1	19	1.91	79.5	21.79	4.0	1.0	34	1	1		1	
17	1	29	1.83	94.1	28.10	4.0	1.0	8	3	3		1	2
21	1	29	1.78	70.2	22.16	5.0	1.5	51	2	1	2	1	2
22	1	18	1.6	62.2	24.30	3.0	6.0	20	4	2	4	1	
23	1	25	1.83	72.3	21.59	3.0	3.0	38	1	1			
24	1	21	1.7	79.2	27.40	3.0	2.0	18	1	1		1	
28	1	25	1.76	75.1	24.24	5.0	1.5	19	1	1		1	
29	1	21	1.84	102.9	30.39	7.0	2.0	45	1	1		1	
31	1	25	1.82	104.1	31.43	5.0	1.5	44	1	1		1	
33	1	22	1.74	63.8	21.07	3.0	2.0	39	1	1		1	
35	1	21	1.88	73.7	20.85	5.0	0.5	32	1	1		2	
37	1	1	1.79	90.8	28.34	4.0	1.0	27	2	1	2	2	
40	1	23	1.94	95.3	25.32	3.0	1.0	46	1	1			
41	1	23	1.7	92.6	32.04	5.0	3.0	36	5	5			
42	1	31	1.76	88.2	28.47	4.0	1.5	13	1			1	
43	1	19	1.72	70.45	23.81	5.0	2.0	49	1	1			
47	1	18	1.759	84.9	27.44	5.0	1.5	25	1			2	1
48	1	18	1.801	69.3	21.37	6.0	2.0	30	1	1		1	
50	1	24	1.778	82	25.94	3.0	1.5	10	2	2		2	1
2	2	20	1.66	77.4	28.09	2.0	2.0	1	1	1		1	
3	2	21	1.57	56.4	22.88	2.0	1.0	9	1	1			
6	2	21	1.76	66.8	21.57	5.0	1.0	12	1	1		1	
7	2	21	1.72	71.2	24.07	5.0	1.0	15	1	1		1	2
8	2	21	1.67	60.8	21.80	2.0	0.5	17	3	3			
10	2	19	1.71	68.8	23.53	7.0	2.0	50	2	2		1	2
11	2	21	1.62	69	26.29	3.0	1.0	2	5	5		2	1
13	2	21	1.64	63.2	23.50	2.0	1.0	42	1			1	2
18	2	21	1.66	77.4	28.09	5.0	1.5	24	1	1		2	1
19	2	20	1.47	43.6	20.18	5.0	1.0	28	1	1		2	
20	2	19	1.52	70.5	30.51	2.0	1.0	22	4	4		4	
25	2	21	1.73	71.6	23.92	2.0	1.5	47	1			1	
26	2	21	1.69	60	21.01	3.0	1.0	14	5	3	5		
27	2	22	1.66	57.8	20.98	3.0	0.5	37	2	2	5		
30	2	25	1.7	57.1	19.76	3.0	2.0	48	1	5		1	
32	2	21	1.79	74.6	23.28	5.0	1.0	35	1	2	1	1	
34	2	22	1.61	56.4	21.76	3.0	3.0	16	1	1		1	
36	2	2	1.66	53.1	19.27	5.0	1.0	41	5	5			
38	2	20	1.54	74	31.20	3.0	1.0	23	1	1			
39	2	22	1.7	65.6	22.70	4.0	2.0	33	1	1		3	
44	2	21	1.704	80.3	27.66	4.0	2.0	31	1	1		2	1
45	2	21	1.705	78.9	27.14	3.0	0.5	29	1	1	2	1	2
46	2	30	1.702	68.4	23.61	4.0	1.0	40	1	1		1	
49	2	20	1.627	53.9	20.36	3.0	1.0	43	1	1	2	2	
51	2	24	1.689	59.2	20.75	2.0	1.0	21	2	2		6	

Frequencies for Activity

	BB<6	VB<6	Gym<6	Plyos<6	MA<6	BB>6	Total
Males	13	3	1	3	1	4	25
Females	13	3	1	3	1	4	25

Appendix I: IRB Approval Form

4/29/2004

THE UNIVERSITY OF NORTH CAROLINA

GREENSBORO

IRB File NUM:

034262

TITLE: Effets of the Abdominal Hollowing Maneuver on Lower Extremity Biomechanics during Drop La

PI: Kulas,Anthony

DEPT: ESS

CO_PIS:

FACULTY SPONSOR: Schmitz,Randy

Action Taken:

_______ eXempt from Full Review

___✓___ **E**xpedited Review

_______ **F**ull IRB Review

Disposition of Application:

___✓___ Approved

_______ **D**isapproved

MODIFICATIONS AND COMMENTS:

APPROVAL DATE*: 5/7/04

EXPIRATION DATE*: 5/7/05

*Approval of Research is for up to **ONE** year only. If your research extends beyond one year, the project must be reviewed before the expiration date prior to continuation.

N:\RSS\apps\uncg\DATA\ORC\facesheet.rpt

Appendix J: Consent Form

THE UNIVERSITY OF NORTH CAROLINA
GREENSBORO

CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: Effects of the Abdominal Hollowing Maneuver on Lower Extremity Biomechanics during Drop Landings

Project Director: Tony Kulas MA, ATC

Participant's Name: ________________

Date of Consent: _____/______/______

DESCRIPTION AND EXPLANATION OF PROCEDURES:
Purpose of the research study: The purpose of this study is to test the effect of performing an abdominal "core stabilization" maneuver on the lower extremity during a drop landing from a box.

This study will examine: The forces on the lower extremity during the drop landings as a result of activating your abdominal muscles.

Participant selection:
By agreeing to participate in the study, you are indicating you are not pregnant, are 18 years of age or older, and have no history of recent surgery, injury or chronic pain in your legs or back and are otherwise healthy.

What you will do in the study:
You will first be asked to give your height, weight, age, and sex. Then we will measure your height and weight on a standard scale with an extension for height measurement. You will have 3 pairs of surface electrodes placed on your abdomen to monitor the activity of your abdominal muscles. You will also have 9 small (1"x1"x1") sensors (2 on the back, 1 on the head, 3 on each leg) placed on your body with double-sided tape or a neoprene sleeve. You will be instructed to drop off a .6 meter (24") box onto two forceplates landing on both feet at the same time. You will be given the opportunity to practice the task until you feel comfortable. When you are ready we will ask you to perform two sets of drop landings (5 landings in the first set, 10 landings in the second) with a 10 minute break in between sessions. After these two sessions are completed data collection for day 1 will be over and you will be asked to report for data collection 24-72 hours (1-3 days) later.

On day two, you will again have the abdominal surface electrodes placed on your abdomen and 9 small (1"x1"x1") sensors (2 on the back, 1 on the head, 3 on each leg) placed on your body with double-sided tape or a neoprene sleeve. You will be instructed to drop off a .6

meter (24") box onto two forceplates landing on both feet at the same time. After this set of 5 drop landings, you will be taught to activate your abdominal muscles that will result in your navel being drawn towards your spine. You will again perform this abdominal maneuver while we monitor the pressure change in the bag. After completion of this abdominal testing, you will again perform 10 drop landings as described above but while simultaneuously performing the abdominal maneuver.

Time required:
2, 120-minute sessions.

Confidentiality:
The information that you give in this study will be handled confidentially. Your information will be assigned a code number. The list connecting your name to this number will be kept in a locked file. When the study is completed and the data have been analyzed, this list will be destroyed. Your name will not be used in any report.

Voluntary participation:
Your participation in the study is completely voluntary, and you may withdraw from the study or ask any questions at any time.

RISKS AND DISCOMFORTS:
There is a slight possibility you may land awkwardly during the testing and subsequently suffer a strain, sprain or contusion. The investigator will be nearby to help protect you from falling. If at any time the testing causes you any discomfort or concern, please notify the investigator immediately. Please contact [redacted] at [redacted] about any research-related injuries.

POTENTIAL BENEFITS:
There are no direct benefits to you for participating in the study.

COMPENSATION/TREATMENT FOR INJURY: (If study poses more than minimal risk, you must include a statement regarding compensation and/or treatment available for injury, and direct participants to contact [redacted] at [redacted] about any research-related injuries they sustain.)
N/A

CONSENT:
By signing this consent form, you agree that you understand the procedures and any risks and benefits involved in this research. You are free to refuse to participate or to withdraw your consent to participate in this research at any time without penalty or prejudice; your participation is entirely voluntary. Your privacy will be protected because you will not be identified by name as a participant in this project.

The research and this consent form have been approved by the University of North Carolina at Greensboro Institutional Review Board, which insures that research involving people follows federal regulations. Questions regarding your rights as a participant in this project can be answered by calling [redacted] (UNCG Research Compliance Officer) at [redacted]. Questions regarding the research itself will be answered by Tony Kulas MA, ATC by calling [redacted]

[redacted] Any new information that develops during the project will be provided to you if the information might affect your willingness to continue participation in the project.

By signing this form, you are agreeing to participate in the project described to you by Tony Kulas.

_____________________________________ ______________

Participant's Signature* Date

Appendix K: Data Sets

SUBJECT	SEX	AGE	HEIGHT	MASS	WEIGHT	BMI	HEAI1	KEAI1	AEAI1
1	1	21	1.783	77.7	761.46	24.4410	-0.3053	-1.48690	-0.8337
2	2	20	1.660	77.4	758.52	28.0883	-0.3292	-0.77057	-0.0117
3	2	21	1.570	56.4	552.72	22.8813	-0.6477	-0.92146	-0.0277
5	1	21	1.840	81.3	796.74	24.0135	-0.1233	-0.80359	-0.0068
6	2	21	1.760	66.8	654.64	21.5651	-0.1787	-3.00625	-2.1359
7	2	21	1.720	71.2	697.76	24.0671	-0.1397	-2.20499	-0.2695
8	2	21	1.670	60.8	595.84	21.8007	-0.4390	-0.78649	-0.0025
9	1	22	1.770	76.8	752.64	24.5140	-0.2370	-0.73805	-0.0244
10	2	19	1.710	68.8	674.24	23.5286	-0.3996	-1.43685	-1.6646
11	2	21	1.620	69.0	676.20	26.2917	-0.5239	-1.18762	-0.0449
12	1	35	1.810	93.3	914.34	28.4790	-0.6020	-2.51208	-2.6242
13	2	21	1.640	63.2	619.36	23.4979	-0.5770	-1.68515	-0.0391
14	1	20	1.820	82.9	812.42	25.0272	-0.1876	-0.78602	-0.0026
15	1	24	1.830	79.1	775.18	23.6197	-0.0721	-0.63006	-0.0220
16	1	19	1.910	79.5	779.10	21.7922	-0.2059	-1.33682	-0.0303
17	1	29	1.830	94.1	922.18	28.0988	-0.3523	-2.85127	-0.7495
18	2	21	1.660	77.4	758.52	28.0883	-0.2826	-1.17941	-0.1108
19	2	20	1.470	43.6	427.28	20.1768	-1.0874	-0.69623	-0.0093
20	2	19	1.520	70.5	690.90	30.5142	-0.6993	-2.26174	-0.0989
21	1	29	1.780	70.2	687.96	22.1563	-0.1025	-3.94556	-0.8101
22	1	18	1.600	62.2	609.56	24.2969	-0.6476	-2.32442	-0.0832
23	1	25	1.830	72.3	708.54	21.5892	-0.1370	-0.91342	-0.4239
24	1	21	1.700	79.2	776.16	27.4048	-0.1593	-2.35985	-1.2013
25	2	21	1.730	71.6	701.68	23.9233	-0.1153	-0.97042	-0.0069
26	2	21	1.690	60.0	588.00	21.0077	-0.6130	-1.70118	-2.4908
27	2	22	1.660	57.8	566.44	20.9755	-0.1543	-1.16120	-0.0163
28	1	25	1.760	75.1	735.98	24.2446	-0.1463	-1.06245	-0.2292
29	1	21	1.840	102.9	1,008.42	30.3934	-0.1805	-4.60662	-1.4118
30	2	25	1.700	57.1	559.58	19.7578	-0.5431	-5.14971	-0.5571
31	1	25	1.820	104.1	1,020.18	31.4274	-0.3795	-3.59164	-4.9311
32	2	21	1.790	74.6	731.08	23.2827	-0.6340	-1.44988	-0.0185
33	1	22	1.740	63.8	625.24	21.0728	-0.2656	-4.03957	-3.6059
34	2	22	1.610	56.4	552.72	21.7584	-0.1473	-3.20796	-2.6692
35	1	21	1.880	73.7	722.26	20.8522	-0.0917	-1.47720	-1.7430
36	2	19	1.660	53.1	520.38	19.2699	-0.5088	-1.16890	-0.4856
37	1	31	1.790	90.8	889.84	28.3387	-0.9153	-1.88263	-2.4187
38	2	20	1.540	74.0	725.20	31.2026	-0.3031	-2.93757	-1.6085
39	2	22	1.700	65.6	642.88	22.6990	-0.2712	-2.05890	-2.3852
40	1	23	1.940	95.3	933.94	25.3215	-0.0634	-4.70530	-1.5395
41	1	23	1.700	92.6	907.48	32.0415	-0.5958	-4.81063	-0.1629
42	1	31	1.760	88.2	864.36	28.4737	-0.2763	-2.51860	-2.6026
43	1	18	1.710	70.4	689.92	24.0758	-0.6442	-4.57541	-3.6484
44	2	21	1.700	80.3	786.94	27.7855	-0.2302	-5.36711	-3.0040
45	2	21	1.710	78.9	773.22	26.9827	-1.2313	-0.84324	-0.0540
46	2	30	1.700	68.4	670.32	23.6678	-0.3435	-1.14445	-2.3294
47	1	18	1.760	84.9	832.02	27.4083	-0.2983	-4.17523	-3.7077
48	1	18	1.800	69.3	679.14	21.3889	-0.8272	-3.95836	-4.9870
49	2	20	1.630	53.9	528.22	20.2868	-0.7445	-1.19021	-1.1209
50	1	24	1.770	93.0	911.40	29.6850	-0.4492	-2.52809	-2.1386
51	2	24	1.690	59.2	580.16	20.7276	-0.4627	-2.07581	-0.8101

SUBJECT	HEAS1	KEAS1	AEAS1	HEAI2	KEAI2	AEAI2	HEAS2	KEAS2
1	-0.0970	-0.3786	-0.1315	-0.1745	-2.4870	-1.3153	-0.1290	-0.8908
2	-0.0323	-0.0354	-0.0008	-0.4085	-0.8022	-0.0553	-0.0334	-0.0477
3	-0.0157	-0.0804	-0.0120	-0.8118	-0.9442	-0.1354	-0.0368	-0.0749
5	-0.0682	-0.0307	-0.0073	-0.2678	-0.8398	-0.0046	-0.0827	-0.0253
6	-0.1067	-0.2584	-0.0020	-0.3833	-2.0780	-2.5770	-0.1039	-0.0539
7	-0.0346	-0.6928	-0.0045	-0.1801	-0.8877	-0.0045	-0.0113	-0.0993
8	-0.0577	-0.0476	-0.0003	-0.5662	-0.8512	-0.0277	-0.0668	-0.0521
9	-0.0216	-0.0217	-0.0005	-0.3164	-0.7957	-0.0558	-0.0032	-0.0326
10	-0.2026	-0.5926	-0.0347	-0.1653	-1.1677	-1.1942	-0.0821	-0.4326
11	-0.2237	-0.0308	-0.0003	-0.6213	-0.5149	-0.0040	-0.1613	-0.0345
12	-0.2472	-0.4790	-0.1416	-0.5788	-1.9896	-1.3051	-0.1536	-0.1133
13	-0.0185	-0.2535	-0.0315	-0.5761	-2.0884	-0.3834	-0.0223	-0.2933
14	-0.0388	-0.0330	-0.0015	-0.1630	-0.8159	-0.0058	-0.0001	-0.1078
15	-0.1056	-0.0802	-0.0001	-0.0597	-0.5421	-0.0120	-0.0543	-0.0123
16	-0.1151	-0.2056	-0.0446	-0.2480	-2.1796	-0.7622	-0.1384	-0.4011
17	-0.1255	-0.1007	-0.0364	-0.4141	-2.3096	-1.3316	-0.1552	-0.1256
18	-0.0301	-0.1242	-0.0127	-0.2841	-0.9887	-0.9368	-0.0583	-0.1136
19	0.0000	-0.0612	-0.0002	-1.4093	-0.7797	-0.0115	-0.0061	-0.1005
20	-0.0402	-0.0348	-0.0011	-0.6559	-1.9730	-0.2320	-0.0522	-0.0480
21	-0.1166	-0.5137	-0.0131	-0.1810	-1.9420	-1.1527	-0.1072	-0.2075
22	-0.0049	-0.2674	-0.0027	-0.8403	-1.5620	-0.0307	-0.0177	-0.0241
23	-0.1829	-0.8239	-0.2857	-0.1158	-0.5686	-0.1518	-0.0705	-0.3258
24	-0.1312	-0.1954	-0.0092	-0.1660	-3.1521	-1.1265	-0.1325	-0.5830
25	-0.1105	-0.0656	-0.0011	-0.1996	-2.4428	-0.4940	-0.1396	-0.5239
26	-0.0488	-0.0863	-0.0011	-0.5512	-1.4086	-0.4838	-0.0192	-0.3630
27	-0.0136	-0.0918	-0.0049	-0.1898	-0.8970	-0.0205	-0.0186	-0.0962
28	-0.0672	-0.1364	-0.0067	-0.1137	-0.4399	-0.0211	-0.1518	-0.0630
29	-0.3952	-0.9051	-0.1231	-0.2398	-4.5948	-1.2290	-0.4121	-1.0440
30	-0.0510	-0.5248	-0.0934	-0.7096	-2.7182	-0.5273	-0.0158	-0.1141
31	-0.1086	-0.9026	-0.4506	-0.2504	-4.2319	-1.8648	-0.0615	-0.8244
32	-0.0233	-0.0881	-0.0003	-0.6104	-1.3551	-0.0089	-0.0225	-0.0706
33	-0.1318	-1.2772	-0.1100	-0.3802	-4.2171	-3.2583	-0.0863	-1.4884
34	-0.0314	-0.5365	-0.0696	-0.1709	-3.0160	-3.4712	-0.0184	-0.5866
35	-0.0244	-0.6625	-0.0222	-0.2206	-2.6112	-1.7402	-0.0342	-0.4292
36	-0.1099	-0.2241	-0.1317	-0.6707	-0.9064	-0.9814	-0.0894	-0.2682
37	-0.2577	-0.3443	-0.0913	-0.9727	-1.4352	-0.9226	-0.3588	-0.2734
38	-0.0927	-0.4252	-0.0588	-0.1938	-2.9925	-0.5320	-0.0292	-0.3804
39	-0.0109	-0.2753	-0.0655	-0.2205	-0.6262	-0.0517	-0.0253	-0.0407
40	-0.0509	-1.4903	-0.1113	-0.0498	-4.6412	-1.4344	-0.0771	-1.3825
41	-0.2125	-1.0528	-0.0277	-0.5302	-4.8155	-0.2386	-0.2028	-1.0658
42	-0.1148	-0.2748	-0.1320	-0.4695	-2.4995	-1.9056	-0.1471	-0.1337
43	-0.2229	-1.0169	-0.1452	-0.6169	-5.4774	-2.2778	-0.2716	-1.0486
44	-0.1769	-1.4247	-0.0461	-0.3617	-4.5048	-6.4971	-0.1831	-1.5384
45	-0.0274	-0.0116	-0.0006	-0.5332	-1.1344	-0.5785	-0.1546	-0.2480
46	-0.1346	-0.2387	-0.0398	-0.1665	-2.6262	-2.9753	-0.0856	-0.2912
47	-0.1285	-1.0819	-0.3541	-0.2158	-3.8171	-5.0641	-0.0969	-1.4150
48	-0.0658	-0.2212	-0.1023	-0.6254	-1.8982	-1.5563	-0.0375	-0.0560
49	-0.0644	-0.0188	-0.0010	-1.4258	-1.9542	-1.4032	-0.0033	-0.0286
50	-0.1737	-0.5952	-0.0848	-0.3852	-2.7007	-0.8723	-0.0893	-0.0280
51	-0.0575	-0.0526	-0.0054	-0.4097	-1.0997	-0.6844	-0.0748	-0.0135

SUBJECT	AEAS2	HEAI3	KEAI3	AEAI3	HEAS3	KEAS3	AEAS3	HEAI4
1	-0.0802	-0.2584	-2.2298	-1.0673	-0.2658	-0.8105	-0.1345	-0.3169
2	-0.0010	-0.4452	-2.7753	-0.4611	-0.1258	-0.2147	-0.0280	-0.4762
3	-0.0078	-0.6267	-0.7422	-0.3329	-0.0158	-0.0808	-0.0112	-0.6552
5	-0.0023	-0.2793	-0.6119	-0.0043	-0.0539	-0.0132	-0.0020	-0.5974
6	-0.0028	-0.0697	-4.4331	-2.1166	-0.1129	-0.6334	-0.0077	-0.1647
7	-0.0034	-0.1348	-1.4638	-0.0379	-0.0072	-0.3417	-0.0042	-0.1460
8	0.0000	-0.5386	-0.6755	-0.0134	-0.0660	-0.0406	-0.0002	-0.5187
9	0.0000	-0.3218	-1.0109	-0.1895	-0.0236	-0.1592	-0.0689	-0.3168
10	-0.0046	-0.0799	-1.8337	-1.8769	-0.0827	-0.6823	-0.0296	-0.0362
11	-0.0005	-0.3432	-0.6239	-0.0020	-0.0747	-0.0434	-0.0001	-0.4714
12	-0.0762	-0.2945	-0.5509	-2.3613	-0.0128	-0.0459	-0.1281	-0.3132
13	-0.0134	-0.5359	-1.7380	-0.2034	-0.2166	-0.2515	-0.0540	-0.9201
14	-0.0010	-0.0758	-0.7248	-0.1036	-0.1173	-0.0067	-0.0060	-0.1106
15	-0.0004	-0.0444	-0.5681	-0.0067	-0.0448	-0.0330	-0.0022	-0.0714
16	-0.1932	-0.3837	-2.7780	-2.6570	-0.2491	-0.9389	-0.3882	-0.4216
17	-0.0175	-0.4049	-1.8784	-1.2075	-0.2025	-0.2131	-0.0265	-0.3706
18	-0.0069	-0.3989	-0.7277	-0.3133	-0.0438	-0.2509	-0.0148	-0.3753
19	-0.0004	-1.0706	-1.0240	-0.0203	-0.0308	-0.0581	-0.0005	-0.6052
20	-0.0013	-0.6419	-3.2063	-2.4263	-0.1549	-0.4230	-0.1625	-0.3047
21	-0.0037	-0.1191	-1.4820	-0.6938	-0.0689	-0.1323	-0.0035	-0.1995
22	-0.0002	-0.7251	-2.6971	-0.4733	-0.0512	-0.2752	-0.0237	-0.6495
23	-0.1056	-0.2024	-0.8841	-0.4465	-0.0332	-0.1653	-0.0258	-0.1502
24	-0.0753	-0.0303	-5.3977	-0.3296	-0.0505	-0.6457	-0.0692	-0.0640
25	-0.0134	-0.1346	-3.2372	-1.0697	-0.1403	-0.4240	-0.0516	-0.3389
26	-0.0326	-0.2542	-3.3397	-2.4392	-0.2147	-0.8343	-0.2841	-0.4850
27	-0.0068	-0.4304	-0.8033	-0.0063	-0.0022	-0.0473	-0.0022	-0.5241
28	-0.0065	-0.0407	-1.4284	-0.4868	-0.2393	-0.7663	-0.0971	-0.1488
29	-0.0593	-0.4952	-4.1352	-1.3477	-0.6097	-0.6864	-0.1017	-0.2632
30	-0.0228	-0.1910	-4.8924	-1.4600	-0.0412	-0.7356	-0.1867	-0.4552
31	-0.2205	-0.0899	-5.3495	-2.0776	-0.0854	-1.0851	-0.0379	-0.1315
32	-0.0003	-0.3925	-3.9848	-2.1479	-0.0529	-0.0951	-0.0264	-0.4096
33	-0.0897	-0.2643	-3.3012	-5.7155	-0.1007	-1.2198	-0.0912	-0.7460
34	-0.1319	-0.1140	-3.8708	-1.8051	-0.0430	-0.5345	-0.2259	-0.0758
35	-0.0370	-0.0847	-3.7297	-2.1415	-0.0150	-1.0972	-0.0396	-0.1207
36	-0.0770	-0.7518	-1.3862	-1.7918	-0.2685	-0.3365	-0.2152	-0.6872
37	-0.0863	-0.3658	-2.3838	-2.0628	-0.2020	-0.5234	-0.1045	-0.2910
38	-0.0436	-0.2044	-1.0347	-0.1725	-0.0751	-0.0517	-0.0016	-0.2899
39	-0.0001	-0.2141	-2.2707	-3.7620	-0.0469	-0.3510	-0.1088	-0.5635
40	-0.1295	-0.4848	-3.8719	-1.9606	-0.1119	-1.1350	-0.1653	-0.4299
41	-0.0151	-0.5769	-3.1514	-1.9121	-0.3703	-0.6910	-0.2138	-0.3907
42	-0.0171	-0.5479	-2.6333	-1.4355	-0.1138	-0.2075	-0.0209	-0.5353
43	-0.1233	-1.0290	-4.0196	-2.7250	-0.3097	-0.6378	-0.1893	-0.7211
44	-0.0016	-0.1646	-4.9534	-4.3046	-0.0376	-1.0991	-0.1499	-0.0919
45	-0.0529	-1.6973	-1.3327	-1.1877	-0.4532	-0.1832	-0.0697	-1.7732
46	-0.0847	-0.1386	-1.9583	-2.4104	-0.0326	-0.1343	-0.0583	-0.4896
47	-0.3126	-0.0686	-3.6038	-4.9229	0.0000	-1.3688	-0.1409	-0.1698
48	-0.0020	-0.9097	-4.0850	-3.5285	-0.1085	-0.1452	-0.0322	-1.7874
49	-0.0005	-0.6318	-0.7734	-1.4402	-0.0036	-0.0380	-0.0012	-0.4462
50	-0.0029	-0.4913	-1.1036	-0.3178	-0.1365	-0.0062	-0.0006	-0.8583
51	-0.0001	-0.6396	-2.7598	-4.1033	-0.1646	-0.2645	-0.0346	-0.8057

SUBJECT	KEAI4	AEAI4	HEAS4	KEAS4	AEAS4	MAXGRF1	COMD1	LSS1
1	-1.8000	-0.7836	-0.2097	-0.5219	-0.1131	1.8402	0.3503	5.3537
2	-1.0809	-0.0456	-0.1333	-0.0424	-0.0005	2.3905	0.3304	7.4477
3	-0.8256	-0.2772	-0.0066	-0.0617	-0.0028	3.1153	0.2674	11.6931
5	-0.9238	-0.0340	-0.0262	-0.0441	-0.0030	5.0781	0.2145	23.7676
6	-6.1001	-2.0227	-0.0464	-1.1472	-0.0726	4.0555	0.3218	12.6561
7	-1.1679	-0.3192	-0.0040	-0.1651	-0.0169	4.0147	0.1894	21.5008
8	-1.0197	-0.0179	-0.0629	-0.0700	-0.0003	2.1882	0.2682	8.3002
9	-0.3637	-0.0535	-0.0080	-0.0073	-0.0001	2.6441	0.2179	12.1609
10	-2.6984	-1.7526	-0.0359	-1.1749	-0.0227	2.2467	0.3879	5.9659
11	-1.6899	-0.0244	-0.0096	-0.0549	0.0000	4.8274	0.2157	22.5646
12	-1.1777	-0.2169	-0.1197	-0.3344	-0.2202	2.9945	0.3055	10.0616
13	-1.0456	-0.0057	-0.0220	-0.0359	-0.0005	3.4021	0.2461	13.8406
14	-0.7627	-0.0107	-0.1044	-0.0011	-0.0065	2.9038	0.1904	15.5100
15	-0.8920	-0.1369	-0.0545	-0.0469	-0.0022	3.7732	0.2581	14.9075
16	-2.9407	-4.0350	-0.1665	-1.3275	-0.5662	4.4047	0.2284	19.3136
17	-0.6612	-0.0657	-0.0204	-0.0251	-0.0005	3.4505	0.4032	8.7476
18	-1.1510	-0.0495	-0.0363	-0.5584	-0.0139	2.2335	0.4201	5.3296
19	-0.8260	-0.0165	-0.0016	-0.1112	-0.0007	4.5238	0.2007	22.6742
20	-3.1774	-4.1932	-0.0703	-0.5072	-0.1252	4.7243	0.3104	15.2972
21	-1.8713	-0.5053	-0.0581	-0.3139	-0.0031	2.4344	0.2961	8.5633
22	-1.5660	-0.0644	-0.0338	-0.2449	-0.0022	3.8094	0.3108	12.4492
23	-1.8881	-0.2281	-0.0298	-0.4676	-0.0953	1.5318	0.4316	3.5798
24	-8.2575	-1.2146	-0.0519	-1.1133	-0.0443	3.0596	0.3060	10.1357
25	-2.6859	-0.8093	-0.1018	-0.4016	-0.0155	3.2377	0.2642	12.3559
26	-2.7648	-3.1798	-0.1568	-0.6245	-0.1400	3.4696	0.3497	9.9296
27	-0.8406	-0.0286	-0.0044	-0.0191	-0.0016	1.9032	0.3524	5.4576
28	-2.4638	-0.7750	-0.1589	-0.7513	-0.0299	2.1388	0.2920	7.4030
29	-2.8920	-0.9245	-0.2543	-0.5109	-0.0984	2.8152	0.2718	10.4135
30	-4.3071	-2.8619	-0.0742	-0.9673	-0.3588	3.3747	0.2657	12.7237
31	-4.3091	-4.5553	-0.0242	-1.0663	-0.2387	3.9033	0.2903	13.6297
32	-3.7088	-2.7771	-0.0492	-0.0433	-0.0036	3.4452	0.3663	9.4928
33	-4.1178	-2.9971	-0.0824	-1.2332	-0.1216	4.4428	0.2419	18.7257
34	-5.1354	-2.2141	-0.0492	-0.5699	-0.1452	3.9559	0.2339	16.9107
35	-1.7061	-0.3176	-0.0212	-0.4037	-0.0163	1.9505	0.2443	8.1067
36	-2.3081	-1.8176	-0.2330	-0.5029	-0.1953	2.5119	0.3212	7.8137
37	-2.8443	-2.1142	-0.1394	-0.6463	-0.0507	2.2596	0.3488	6.5620
38	-1.1458	-0.0903	-0.3683	-0.0675	-0.0010	3.6356	0.2628	14.6554
39	-1.5804	-2.2800	-0.0609	-0.1129	-0.1833	2.9249	0.2561	11.6267
40	-4.5052	-1.5131	-0.0709	-1.5581	-0.1925	2.2495	0.3641	6.1912
41	-2.7410	-2.5866	-0.2827	-0.5257	-0.2100	2.8814	0.4024	7.1788
42	-1.0733	-0.8174	-0.0866	-0.0331	-0.0006	3.5618	0.2406	15.0453
43	-4.8276	-2.3025	-0.1047	-1.0901	-0.0927	3.4836	0.3255	10.7697
44	-5.5475	-2.2972	-0.1095	-1.3082	-0.1185	4.9630	0.2383	20.8569
45	-1.2390	-1.1077	-0.1420	-0.0834	-0.0300	2.8800	0.3882	7.4544
46	-1.8504	-1.0151	-0.0431	-0.0426	0.0000	3.7770	0.2691	14.1453
47	-3.7218	-5.5131	-0.0188	-1.5075	-0.0833	3.4932	0.2510	14.2085
48	-3.3126	-4.7152	-0.4627	-0.4079	-0.0603	3.7971	0.2555	14.8986
49	-1.1133	-0.5864	-0.0163	-0.0134	-0.0006	3.2489	0.3656	9.0466
50	-1.9676	-0.7413	-0.0707	-0.0167	-0.0005	2.9120	0.4350	6.7817
51	-2.8251	-3.6903	-0.0497	-0.3114	-0.0383	4.1046	0.3212	12.9141

SUBJECT	MAXGRF2	COMD2	LSS2	MAXGRF3	COMD3	LSS3	MAXGRF4	COMD4
1	1.5230	0.3874	3.9318	1.5058	0.3962	3.8049	1.6942	0.3610
2	2.3966	0.3640	6.6085	3.0652	0.3782	8.2279	2.5674	0.3466
3	3.5576	0.2701	13.1895	3.6083	0.2797	12.9937	3.8610	0.2919
5	4.8156	0.2374	20.3725	4.5526	0.2371	19.2555	4.6813	0.2343
6	3.7665	0.3326	11.4710	3.7862	0.3202	11.9801	3.7667	0.3249
7	2.8174	0.2264	12.4782	2.5702	0.1995	12.9182	2.5571	0.2168
8	2.5722	0.2825	9.1599	2.8262	0.2810	10.0592	3.0358	0.2897
9	2.4553	0.2284	10.7704	2.2241	0.2607	8.5649	2.2531	0.2473
10	1.9636	0.3910	5.0570	2.0894	0.3674	5.7244	2.4595	0.3535
11	4.6797	0.2159	21.7647	4.4791	0.2191	20.7805	4.1052	0.2292
12	2.7607	0.3290	8.5451	3.0369	0.3202	9.5322	3.5960	0.3577
13	3.2281	0.2697	12.0341	3.1836	0.2513	12.7577	3.2306	0.2655
14	2.7095	0.1901	14.4334	2.9637	0.1990	14.9207	2.8343	0.2001
15	2.5937	0.3271	8.0643	2.5856	0.2863	9.0445	3.0252	0.2688
16	4.2110	0.2399	17.7837	3.6567	0.2863	12.7993	3.7442	0.2709
17	3.3825	0.4093	8.3160	3.2660	0.4272	7.8136	3.4833	0.3150
18	1.9045	0.4425	4.3150	1.7397	0.4659	3.7947	1.9445	0.4652
19	4.0528	0.2203	18.4229	4.6496	0.2271	20.4738	4.5463	0.2259
20	4.1637	0.3379	12.3829	4.8312	0.3109	15.6354	4.0248	0.3268
21	1.9681	0.3205	6.1623	2.0414	0.3089	6.6837	2.0408	0.3053
22	3.2561	0.3760	8.7070	3.0826	0.3871	7.9710	3.1745	0.3342
23	1.5176	0.4041	3.7606	1.5388	0.4713	3.2747	1.5059	0.4492
24	2.6168	0.3395	7.7179	2.4724	0.3269	7.6171	3.0932	0.2934
25	3.4868	0.2898	12.0865	3.7684	0.2713	13.9747	3.8066	0.2747
26	3.0581	0.3756	8.1419	3.4549	0.3898	8.8849	3.0611	0.4172
27	1.6884	0.3838	4.4060	1.6329	0.3439	4.7580	1.4837	0.3677
28	2.2407	0.2833	7.9397	1.7314	0.3243	5.3996	2.3435	0.3215
29	2.5477	0.2990	8.6859	2.5000	0.3539	7.2334	2.4211	0.3264
30	3.3558	0.2867	11.8221	3.3735	0.2957	11.5216	3.3576	0.3207
31	2.9868	0.3042	9.9647	3.5219	0.2874	12.2624	3.7240	0.2405
32	3.0951	0.3934	7.8924	3.5380	0.3649	9.8560	3.4605	0.3583
33	4.4874	0.2934	15.4963	4.3101	0.2885	15.0373	5.2829	0.2685
34	3.5439	0.2362	15.0626	3.3285	0.2502	13.4200	3.1567	0.2732
35	1.8333	0.2689	6.8618	1.9347	0.2943	6.7503	1.8058	0.3238
36	2.5255	0.3418	7.4927	2.7466	0.3476	7.8152	3.5883	0.3756
37	1.8950	0.3791	4.9822	2.2626	0.4847	4.6906	2.8049	0.4484
38	2.6692	0.2620	10.2608	3.2841	0.2287	14.3534	4.1712	0.3238
39	3.1048	0.2580	12.0538	3.9284	0.2680	15.1198	4.2338	0.2775
40	1.8747	0.4394	4.3961	2.1366	0.4257	5.0757	2.1873	0.4131
41	2.6323	0.4065	6.5814	2.8863	0.4384	6.5998	2.7892	0.3997
42	3.8386	0.2589	14.9034	3.3656	0.2773	12.1581	3.7394	0.3016
43	3.6218	0.4062	8.9344	3.4092	0.4037	8.5566	3.4151	0.3789
44	4.4886	0.2456	18.3057	4.7079	0.2527	18.6245	4.3881	0.2388
45	2.2886	0.4112	5.6131	3.1427	0.3734	8.5467	2.2092	0.4864
46	3.4814	0.2799	12.4596	2.9835	0.2880	10.5194	3.3895	0.2876
47	3.3215	0.2608	12.8169	3.5110	0.2473	14.2128	3.1666	0.2554
48	3.5729	0.2482	14.4322	2.7653	0.3067	9.6019	2.2562	0.3390
49	3.1868	0.3883	8.3640	2.7326	0.3483	7.9085	2.6101	0.3500
50	2.9163	0.4303	6.8473	2.6470	0.4258	6.2829	3.0209	0.3931
51	3.1192	0.3515	8.8688	3.1879	0.3547	9.1005	2.9942	0.3636

SUBJECT	LSS4	RAPRA1	EOPRA1	TRAPRA1	RAIRA1	EOIRA1	TRAIRA1	RAPRA2
1	4.7155	0.3222	0.1141	0.5637	0.0463	0.0968	0.8569	0.3578
2	7.5328	0.3391	0.2490	0.4120	0.1788	0.5238	0.2973	0.3067
3	13.3262	0.2357	0.2109	0.5534	0.3665	0.2851	0.3484	0.2558
5	20.0319	0.0870	0.1325	0.7805	0.1218	0.1574	0.7208	0.0607
6	11.7048	0.1546	0.3651	0.4803	0.2184	0.4944	0.2872	0.1999
7	11.9096	0.2043	0.2812	0.5145	0.2156	0.4529	0.3315	0.1926
8	10.4907	0.0092	0.0291	0.9618	0.0042	0.0444	0.9514	0.0084
9	9.1448	0.1291	0.1379	0.7330	0.1656	0.1954	0.6389	0.1628
10	7.0809	0.2293	0.4195	0.3512	0.2615	0.4151	0.3235	0.1888
11	18.1277	0.1717	0.3139	0.5143	0.4680	0.3677	0.1643	0.1814
12	11.0899	0.1561	0.1515	0.6925	0.1099	0.1250	0.7651	0.1981
13	12.1646	0.2221	0.2778	0.5000	0.5264	0.2518	0.2217	0.1684
14	14.2054	0.1441	0.1536	0.7023	0.0849	0.1667	0.7483	0.1563
15	11.3163	0.4060	0.1155	0.4786	0.1221	0.6199	0.2580	0.3662
16	13.9339	0.4332	0.0719	0.4949	0.6327	0.0578	0.3095	0.2491
17	11.3774	0.2138	0.1418	0.6444	0.1136	0.1393	0.7471	0.3909
18	4.1850	0.1820	0.1146	0.7035	0.4484	0.1463	0.4053	0.2216
19	20.1424	0.3365	0.2446	0.4189	0.2725	0.2585	0.4690	0.3287
20	12.3478	0.4792	0.1846	0.3362	0.8478	0.1066	0.0456	0.4863
21	6.7363	0.1299	0.1243	0.7458	0.0437	0.1215	0.8348	0.1485
22	9.5445	0.2237	0.1478	0.6285	0.0895	0.2167	0.6938	0.2470
23	3.3655	0.2277	0.1423	0.6300	0.0438	0.1190	0.8372	0.2249
24	10.6153	0.2547	0.2576	0.4877	0.2000	0.3118	0.4882	0.2256
25	13.9489	0.1915	0.1898	0.6187	0.3222	0.4112	0.2666	0.1904
26	7.3515	0.3866	0.3076	0.3057	0.4105	0.4306	0.1589	0.4375
27	4.0418	0.1699	0.1686	0.6615	0.1153	0.3291	0.5556	0.1535
28	7.2801	0.0371	0.1326	0.8303	0.0635	0.1842	0.7523	0.0278
29	7.5358	0.2081	0.1145	0.6774	0.3053	0.1394	0.5553	0.2030
30	10.7973	0.1192	0.3801	0.5007	0.0585	0.4373	0.5042	0.1289
31	15.6192	0.1574	0.4881	0.3545	0.1253	0.4250	0.4497	0.1082
32	9.6608	0.3584	0.1974	0.4442	0.2208	0.3413	0.4379	0.4144
33	20.0693	0.1198	0.0978	0.7824	0.0636	0.1534	0.7829	0.0948
34	11.6292	0.2321	0.2149	0.5529	0.1962	0.4719	0.3318	0.2218
35	5.5718	0.1387	0.1757	0.6856	0.0674	0.1272	0.8054	0.1918
36	9.6304	0.1318	0.1870	0.6812	0.0558	0.4683	0.4759	0.1045
37	6.3910	0.2740	0.2517	0.4742	0.5311	0.1368	0.3322	0.2163
38	14.3308	0.3080	0.2378	0.4542	0.6066	0.2109	0.1824	0.2945
39	15.3665	0.3952	0.2106	0.3942	0.2639	0.2869	0.4491	0.3892
40	5.3715	0.2333	0.2225	0.5442	0.1321	0.2357	0.6321	0.2762
41	7.0636	0.2635	0.2674	0.4691	0.2938	0.5401	0.1661	0.3095
42	12.5197	0.1163	0.1866	0.6970	0.3216	0.3200	0.3583	0.1292
43	9.0511	0.1728	0.1883	0.6389	0.1262	0.1990	0.6748	0.2387
44	18.4113	0.4443	0.1934	0.3623	0.4394	0.3036	0.2569	0.4320
45	4.6751	0.1783	0.0780	0.7437	0.1445	0.0639	0.7916	0.2111
46	11.8113	0.2347	0.1211	0.6442	0.3390	0.1516	0.5094	0.2810
47	12.4584	0.2677	0.1680	0.5643	0.0732	0.5778	0.3491	0.2161
48	6.7295	0.0671	0.2178	0.7151	0.0238	0.3479	0.6283	0.0872
49	7.6719	0.1961	0.2187	0.5852	0.2000	0.1634	0.6366	0.2177
50	8.0045	0.1925	0.2979	0.5095	0.0641	0.3235	0.6123	0.1812
51	8.2744	0.1568	0.2628	0.5804	0.1271	0.3771	0.4959	0.1757

SUBJECT	EOPRA2	TRAPRA2	RAIRA2	EOIRA2	TRAIRA2	RAPRA3	EOPRA3	TRAPRA3
1	0.0894	0.5529	0.0566	0.1086	0.8348	0.2634	0.0960	0.6406
2	0.2411	0.4523	0.2275	0.4405	0.3320	0.2036	0.3717	0.4247
3	0.2150	0.5292	0.4292	0.2755	0.2953	0.1942	0.2333	0.5725
5	0.1226	0.8167	0.1076	0.1203	0.7721	0.0683	0.1414	0.7904
6	0.2179	0.5822	0.3386	0.2666	0.3948	0.1760	0.1760	0.6479
7	0.2666	0.5408	0.2098	0.4271	0.3632	0.1565	0.3142	0.5293
8	0.0280	0.9636	0.0035	0.0266	0.9699	0.1599	0.2773	0.5628
9	0.1590	0.6782	0.1453	0.2659	0.5888	0.2506	0.0946	0.6548
10	0.4791	0.3321	0.1788	0.5750	0.2461	0.2884	0.2453	0.4663
11	0.2920	0.5266	0.4964	0.3787	0.1250	0.1739	0.3068	0.5193
12	0.1771	0.6248	0.0920	0.1075	0.8005	0.2864	0.2048	0.5088
13	0.1913	0.6403	0.4571	0.1352	0.4077	0.1655	0.2002	0.6343
14	0.1166	0.7271	0.1058	0.1826	0.7116	0.2107	0.2087	0.5806
15	0.1398	0.4940	0.2295	0.3846	0.3859	0.2251	0.1103	0.6646
16	0.0892	0.6617	0.2094	0.1057	0.6849	0.1876	0.1105	0.7019
17	0.2562	0.3529	0.2451	0.2600	0.4949	0.2505	0.2637	0.4858
18	0.0916	0.6868	0.4448	0.1163	0.4389	0.1342	0.1397	0.7260
19	0.2557	0.4157	0.2587	0.2562	0.4851	0.2495	0.2681	0.4824
20	0.1742	0.3395	0.8983	0.0593	0.0424	0.4102	0.2558	0.3340
21	0.0939	0.7576	0.0376	0.1313	0.8311	0.0732	0.0887	0.8381
22	0.1277	0.6253	0.1055	0.2892	0.6053	0.2038	0.0845	0.7117
23	0.1269	0.6483	0.0340	0.1287	0.8373	0.2335	0.1393	0.6272
24	0.2530	0.5214	0.1641	0.2022	0.6337	0.1937	0.3315	0.4748
25	0.2193	0.5903	0.3045	0.4235	0.2720	0.1528	0.1721	0.6751
26	0.2837	0.2789	0.2695	0.5469	0.1836	0.0629	0.0813	0.8558
27	0.1743	0.6722	0.1286	0.2918	0.5796	0.1704	0.2410	0.5887
28	0.1280	0.8442	0.0510	0.1629	0.7861	0.0957	0.4529	0.4514
29	0.1324	0.6647	0.2369	0.0669	0.6961	0.2314	0.0880	0.6806
30	0.3319	0.5392	0.0521	0.4304	0.5175	0.1192	0.3801	0.5007
31	0.5083	0.3835	0.1289	0.5497	0.3214	0.1553	0.2658	0.5789
32	0.1889	0.3967	0.1883	0.3437	0.4680	0.2539	0.1840	0.5620
33	0.0889	0.8163	0.0358	0.1596	0.8046	0.1827	0.1446	0.6728
34	0.2703	0.5079	0.1774	0.5385	0.2841	0.2802	0.1694	0.5503
35	0.2088	0.5994	0.0830	0.2066	0.7104	0.1815	0.2746	0.5439
36	0.2179	0.6776	0.0661	0.4778	0.4561	0.1367	0.2617	0.6016
37	0.2563	0.5274	0.3299	0.2556	0.4145	0.3098	0.4036	0.2866
38	0.2184	0.4871	0.4044	0.3670	0.2286	0.3231	0.2468	0.4302
39	0.1886	0.4222	0.2089	0.3206	0.4704	0.4036	0.2714	0.3250
40	0.2448	0.4789	0.1782	0.2690	0.5528	0.0610	0.0626	0.8764
41	0.2242	0.4663	0.5442	0.2711	0.1847	0.1572	0.2564	0.5865
42	0.2002	0.6706	0.1494	0.4416	0.4090	0.1155	0.1990	0.6855
43	0.1883	0.5730	0.0856	0.1782	0.7361	0.3840	0.3168	0.2992
44	0.1945	0.3735	0.4077	0.3954	0.1970	0.4196	0.2093	0.3711
45	0.0877	0.7012	0.1494	0.0631	0.7876	0.4244	0.1802	0.3954
46	0.1318	0.5873	0.3310	0.1678	0.5012	0.2151	0.1553	0.6295
47	0.1904	0.5935	0.1766	0.3538	0.4697	0.2152	0.1740	0.6108
48	0.2396	0.6732	0.0369	0.3192	0.6439	0.1155	0.3385	0.5459
49	0.1425	0.6398	0.2446	0.1582	0.5972	0.1298	0.2200	0.6502
50	0.3313	0.4875	0.0405	0.2116	0.7479	0.2082	0.2992	0.4926
51	0.2665	0.5578	0.1283	0.3111	0.5606	0.2309	0.2716	0.4975

SUBJECT	RAIRA3	EOIRA3	TRAIRA3	RAPRA4	EOPRA4	TRAPRA4	RAIRA4	EOIRA4
1	0.0495	0.0980	0.8525	0.2364	0.0943	0.6692	0.0371	0.0900
2	0.1435	0.6459	0.2106	0.1747	0.3687	0.4566	0.1059	0.6212
3	0.1881	0.2467	0.5652	0.1646	0.1938	0.6415	0.2419	0.2673
5	0.0750	0.2984	0.6266	0.0567	0.1236	0.8197	0.1513	0.3179
6	0.3264	0.1950	0.4787	0.1654	0.1635	0.6711	0.2478	0.2354
7	0.1930	0.4143	0.3927	0.1209	0.2714	0.6077	0.1872	0.4960
8	0.1290	0.3052	0.5658	0.1364	0.2057	0.6579	0.1074	0.2399
9	0.2716	0.1365	0.5919	0.2001	0.0648	0.7350	0.2400	0.0968
10	0.2661	0.1131	0.6208	0.2504	0.2643	0.4853	0.3350	0.1722
11	0.3005	0.4512	0.2483	0.1694	0.2479	0.5827	0.2107	0.5940
12	0.2345	0.1235	0.6420	0.1285	0.1253	0.7461	0.1380	0.1272
13	0.4767	0.1477	0.3757	0.1514	0.1838	0.6648	0.4176	0.1742
14	0.0987	0.2368	0.6645	0.0357	0.0795	0.8849	0.0289	0.1070
15	0.1056	0.2566	0.6378	0.1823	0.1028	0.7150	0.1151	0.1981
16	0.0633	0.1028	0.8339	0.1238	0.1175	0.7587	0.0538	0.1182
17	0.2029	0.2300	0.5671	0.2391	0.1545	0.6064	0.1658	0.1530
18	0.1537	0.4031	0.4432	0.1172	0.1053	0.7775	0.2037	0.1949
19	0.2422	0.2684	0.4894	0.2488	0.2839	0.4673	0.2354	0.3098
20	0.6967	0.2362	0.0671	0.2695	0.2800	0.4505	0.3157	0.4812
21	0.0199	0.0787	0.9014	0.0683	0.0426	0.8891	0.0209	0.0438
22	0.1336	0.1429	0.7234	0.1482	0.1446	0.7072	0.0941	0.1398
23	0.0651	0.3177	0.6172	0.1288	0.1093	0.7619	0.0448	0.3577
24	0.1873	0.2947	0.5181	0.1809	0.2012	0.6179	0.1709	0.3207
25	0.2719	0.3125	0.4155	0.0786	0.1566	0.7648	0.1651	0.4151
26	0.0723	0.3698	0.5579	0.0794	0.0669	0.8537	0.0512	0.2719
27	0.1761	0.3813	0.4425	0.1008	0.1814	0.7178	0.1404	0.3184
28	0.3077	0.4337	0.2586	0.0771	0.5015	0.4214	0.1336	0.6137
29	0.1092	0.3446	0.5461	0.1287	0.0647	0.8066	0.1197	0.2279
30	0.0585	0.4373	0.5042	0.1507	0.2814	0.5679	0.0803	0.3802
31	0.3820	0.4097	0.2083	0.1119	0.1943	0.6939	0.1458	0.3921
32	0.1647	0.2101	0.6251	0.1996	0.1880	0.6124	0.1166	0.2089
33	0.0929	0.1717	0.7354	0.1048	0.1506	0.7446	0.0774	0.1688
34	0.2701	0.4050	0.3249	0.1617	0.1811	0.6572	0.1511	0.4049
35	0.0739	0.2444	0.6817	0.1087	0.2160	0.6753	0.0321	0.2172
36	0.0648	0.4002	0.5350	0.0628	0.1669	0.7703	0.0511	0.2955
37	0.1893	0.5553	0.2554	0.2245	0.2301	0.5454	0.1886	0.2627
38	0.4766	0.2895	0.2339	0.2887	0.2358	0.4755	0.4722	0.3182
39	0.3396	0.4068	0.2536	0.2589	0.2778	0.4633	0.1979	0.4055
40	0.0225	0.0599	0.9175	0.0506	0.0476	0.9018	0.0178	0.0521
41	0.2480	0.4910	0.2609	0.1288	0.1397	0.7316	0.2133	0.2392
42	0.0925	0.4972	0.4103	0.0884	0.1763	0.7352	0.1178	0.2898
43	0.1130	0.5022	0.3848	0.1357	0.1912	0.6731	0.0464	0.2442
44	0.6603	0.1457	0.1940	0.2826	0.2377	0.4797	0.6130	0.1969
45	0.3426	0.2456	0.4117	0.3985	0.1821	0.4193	0.4090	0.2107
46	0.2694	0.2199	0.5107	0.1126	0.1545	0.7329	0.1334	0.1989
47	0.0800	0.4560	0.4640	0.1118	0.1693	0.7189	0.0632	0.4680
48	0.0379	0.3078	0.6544	0.1079	0.2607	0.6314	0.0195	0.2055
49	0.0944	0.1013	0.8043	0.0955	0.1951	0.7094	0.1309	0.1096
50	0.1244	0.1839	0.6917	0.1879	0.2492	0.5629	0.1489	0.2192
51	0.1515	0.3664	0.4820	0.1087	0.1809	0.7104	0.1132	0.2905

SUBJECT	TRAIRA4	TRA4PTRA	TRASTRA	HTJD1	KTJD1	ATJD1	HTJD2	KTJD2
1	0.8729	0.8603	0.9535	51.6150	72.6900	67.5900	51.5184	78.9399
2	0.2729	0.6199	0.5729	31.2981	76.6864	82.6971	39.1715	79.9134
3	0.4908	0.5879	0.8647	29.4301	64.6768	63.9432	31.1729	64.2957
5	0.5308	0.9407	0.8518	12.7551	61.4950	66.2025	18.7770	63.9989
6	0.5168	0.7884	0.6617	22.1890	64.8689	69.5137	28.4581	66.4400
7	0.3168	0.7652	0.9136	10.0385	53.2655	66.8343	13.4754	60.0684
8	0.6527	0.8251	0.7936	35.7498	74.9769	73.6583	30.3399	78.4174
9	0.6632	0.8383	0.8658	13.9304	47.1994	57.9851	12.2114	48.8286
10	0.4928	0.9403	0.6667	54.9756	74.5306	70.1902	48.3134	73.8987
11	0.1953	0.3331	0.3767	18.3056	55.7556	65.2547	27.4254	54.9157
12	0.7348	0.9039	0.8249	51.3067	65.2291	58.0215	50.0108	71.5226
13	0.4082	0.6081	0.7729	37.4815	69.5684	62.3254	42.7163	72.4182
14	0.8641	0.5706	0.8809	9.6893	45.8276	59.8731	10.1184	42.9067
15	0.6869	0.8625	0.8838	12.5584	57.6349	70.5759	18.0129	65.2413
16	0.8280	0.7462	0.9041	7.3787	45.8325	64.4409	11.6888	50.5553
17	0.6812	0.6403	0.6874	60.6693	82.6493	66.9644	56.7919	82.1914
18	0.6013	0.6524	0.8029	59.1204	87.8458	68.1398	62.6612	92.7666
19	0.4549	0.4284	0.1380	33.0856	55.5476	58.4494	38.9272	65.9783
20	0.2032	0.7823	0.7824	35.8027	76.0576	73.4045	42.8051	78.9274
21	0.9353	0.8917	0.9744	26.6870	71.0681	63.9962	28.0181	76.8602
22	0.7661	0.7571	0.8626	47.1416	77.7569	65.2722	59.9142	86.2104
23	0.5975	0.6115	0.9036	51.6065	92.7901	73.5105	49.5037	91.0014
24	0.5084	0.6267	0.6785	39.0852	63.0921	61.3382	44.4665	66.7304
25	0.4198	0.7918	0.8651	17.2709	67.8366	74.1739	21.7105	68.5389
26	0.6769	0.9254	0.7764	40.1393	74.3570	74.3969	40.5532	74.9911
27	0.5412	0.8547	0.8178	44.4982	82.2784	58.6134	49.5645	90.5105
28	0.2527	0.4568	0.2946	36.8882	59.7777	60.6019	39.0734	60.9901
29	0.6524	0.9869	0.9162	35.3374	67.3004	50.3343	34.0848	71.9176
30	0.5395	0.9349	0.6796	28.2354	58.7885	58.5341	27.2779	62.1812
31	0.4621	0.8779	0.5441	43.7498	62.9815	65.3370	48.9946	65.4171
32	0.6745	0.8961	0.9580	34.1340	86.1282	78.1309	35.3650	88.5706
33	0.7538	0.9135	0.8198	22.2646	66.1736	72.7212	28.0948	74.9823
34	0.4440	0.4825	0.9456	12.8428	51.8024	64.9275	15.4770	55.3177
35	0.7507	0.9956	0.8904	18.1410	60.0085	71.6458	18.6169	68.6718
36	0.6534	0.8260	0.9494	43.8284	72.3969	60.4697	46.0099	76.1632
37	0.5488	0.2562	0.8938	55.5033	73.4139	48.9955	62.1395	65.9941
38	0.2096	0.5067	0.5302	26.1524	61.6882	65.4648	30.5387	69.9988
39	0.3966	0.8275	0.7395	23.8461	62.8344	71.8091	25.6852	64.4293
40	0.9301	0.9780	0.9678	36.0618	76.6889	69.3104	46.1996	88.1071
41	0.5475	0.9570	0.7591	62.9476	79.8073	60.3725	60.8601	77.2918
42	0.5925	0.8354	0.8654	20.0396	62.9894	66.5152	32.6733	64.0578
43	0.7094	0.4423	0.4295	39.1979	74.8185	68.5282	47.5610	83.7626
44	0.1901	0.7552	0.5340	8.8748	57.3326	71.1116	9.2635	63.9834
45	0.3803	0.6083	0.8246	66.4741	72.6263	61.7386	64.5744	70.7185
46	0.6677	0.7612	0.5976	24.4381	58.3360	66.5713	29.2375	57.5197
47	0.4688	0.4896	0.5306	27.3539	63.0403	63.1108	24.4388	67.4598
48	0.7750	0.9412	0.8766	37.4913	71.1882	59.9948	34.9146	67.8694
49	0.7595	0.8522	0.8531	55.2127	86.2696	72.0916	62.2678	91.6553
50	0.6318	0.4296	0.8211	53.1950	94.4744	79.2102	50.0283	94.1987
51	0.5964	0.7470	0.5788	38.5566	75.3039	68.9830	43.7143	78.4854

SUBJECT	ATJD2	HTJD3	KTJD3	ATJD3	HTJD4	KTJD4	ATJD4
1	68.7728	54.8242	76.2596	63.5490	47.3565	74.7491	59.8508
2	79.6314	50.4080	78.0065	75.0396	34.2179	74.8624	77.7595
3	64.3020	28.3190	65.4354	62.0996	28.6090	62.4870	62.3851
5	66.6340	21.4751	63.0514	63.3508	19.6913	62.6494	64.2683
6	71.5600	26.8607	65.4035	70.2002	31.5046	71.1056	66.4274
7	66.7511	11.3751	52.7236	68.4739	13.9400	57.7241	68.8313
8	76.4127	28.7894	76.1214	71.8168	28.9652	81.5640	75.9925
9	57.7302	21.6476	58.6348	56.0733	19.1238	57.0866	55.9660
10	70.5978	44.2401	71.6293	69.7396	40.6051	74.8482	71.4664
11	60.4987	22.3477	54.3207	59.7694	28.5866	60.3876	61.5125
12	60.2024	24.4977	33.2105	60.3163	35.2891	64.1822	58.4444
13	61.6388	47.9777	68.5800	62.3442	46.7745	72.9835	59.6593
14	60.0384	10.8825	40.7902	61.7146	6.6551	47.9912	66.2464
15	69.7523	15.6292	65.8920	63.5484	12.3114	64.3130	65.2676
16	65.0928	26.1698	64.4878	63.6403	16.5088	58.9028	68.1734
17	67.0376	68.2928	82.6128	67.5986	50.6204	69.6597	63.6353
18	66.2344	71.0852	100.5896	70.0838	58.2248	104.7525	69.7802
19	58.4046	33.0432	68.1922	66.0644	31.1453	68.4851	65.9651
20	72.0893	38.8012	74.0688	67.4283	37.7511	80.3395	74.8462
21	61.9665	33.6189	79.8432	62.1012	35.5459	76.1600	63.8020
22	73.9054	56.0583	84.0645	65.8394	43.1802	79.9242	71.6663
23	72.1602	60.2863	98.7872	71.9636	51.1573	95.2131	73.7906
24	56.0471	41.3543	67.5441	55.9614	26.4410	66.7750	64.5506
25	71.8408	23.5751	66.9118	69.4930	16.3723	71.3499	75.1212
26	75.6072	49.5517	80.3975	69.2572	57.9421	80.5094	73.0971
27	55.8995	50.8509	76.2525	57.5236	54.3665	83.3141	57.6100
28	57.8950	47.7311	65.7445	57.3905	51.2368	65.6900	57.0288
29	51.1112	55.1463	77.2837	47.0231	26.9164	73.9430	48.4119
30	62.3471	26.5591	63.6973	60.2727	23.5218	63.5490	67.2544
31	63.4548	33.5940	65.0743	58.5833	22.6015	55.4863	65.1636
32	81.1027	41.4089	81.7422	76.4646	34.5345	79.7954	81.6537
33	71.6124	26.6215	76.9047	64.7768	34.3381	74.1599	63.2514
34	64.9302	22.1696	55.3901	65.7578	22.3351	60.1069	66.1269
35	66.9671	25.1135	69.6482	62.6031	35.0388	72.0708	62.0863
36	52.2147	48.7678	71.4011	53.3925	53.3573	86.6382	67.2402
37	37.1656	66.5133	79.2284	52.6625	55.6776	79.3827	52.1030
38	63.0856	30.4500	63.2874	66.0303	29.7997	65.0374	65.7844
39	74.0635	27.2483	64.0597	71.1104	30.9010	68.2776	74.7340
40	70.2287	53.1355	78.4501	67.2054	47.9551	79.4745	67.4562
41	61.0622	70.5918	83.2409	60.5425	54.9854	78.2956	62.5088
42	65.5949	42.6120	68.4355	62.5804	45.0088	72.5427	66.1371
43	67.0714	58.5735	83.2986	69.1829	45.2918	80.7054	68.5830
44	73.0719	23.4474	60.8583	68.2229	19.9010	61.2992	71.5690
45	62.7371	72.5288	62.1999	61.0724	76.9315	75.3987	63.5560
46	66.1463	34.5747	56.7597	63.0949	33.0318	58.2936	70.7407
47	65.2935	27.1129	62.3704	65.2219	22.3509	65.9875	63.9201
48	58.0149	48.3111	73.3218	62.6341	54.6507	69.6394	67.1855
49	75.4166	53.2059	85.2052	67.1835	44.6972	87.2931	74.1206
50	79.8238	54.5313	88.6582	72.6791	54.9025	85.1363	77.0431
51	72.4974	49.5218	80.8953	71.6038	46.3429	84.6089	73.2305

*Subject #4 was removed from the study due to technical data acquisition problems of day 2

Key for Data Sheets

SUBJECT	Subject number
SEX	1=males, 2=females
AGE	Unit = years
HEIGHT	Unit = meters
MASS	Unit = killograms
WEIGHT	Unit = newtons
BMI	Unit = mass/(height squared)
HEAI	Hip Energy Absorption Impact Phase
KEAI	Knee Energy Absorption Impact Phase
AEAI	Ankle Energy Absorption Impact Phase
HEAS	Hip Energy Absorption Stabilization Phase
KEAS	Knee Energy Absorption Stabilization Phase
AEAS	Ankle Energy Absorption Stabilization Phase
MAXGRF	Maximum Ground Reaction Force
COMD	Body's Center of Mass Displacement
LSS	Leg Spring Stiffness
RAPRA	Rectus Abdominis Preactivation Ratio
EOPRA	External Oblique Preactivation Ratio
TRAPRA	Transversus Abdominis-Internal Oblique Preactivation Ratio
RAIRA	Rectus Abdominis Post-Impact Ratio
EOIRA	External Oblique Post-Impact Ratio
TRAIRA	Transversus Abdominis-Internal Oblique Post-Impact Ratio
TRA4PTRA	Transversus Abdominis-Internal Oblique Ratio during Abdominal Hollowing in Four Point Kneeling
TRASTRA	Transversus Abdominis-Internal Oblique Ratio during Abdominal Hollowing in Standing
HTJD	Hip Total Joint Displacement
KTJD	Knee Total Joint Displacement
ATJD	Ankle Total Joint Displacement

*The number to the right of all labels in the data set corresponds to the condition. Conditions 1 & 2 correspond to first and second control conditions on day 1. Conditions 3 & 4 correspond to control and experimental conditions on day 2 respectively